Neuromechanics of movement in lower limb amputees

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Dissertation University of Groningen, The Netherlands — with references — with summary in Dutch and German

ISBN: 978-90-367-5321-0
ISBN: 978-90-367-5320-3 (electronic version)

rijksuniversiteit
groningen

Neuromechanics of Movement in Lower Limb Amputees

Proefschrift

ter verkrijging van het doctoraat in de
Medische Wetenschappen
aan de Rijksuniversiteit Groningen
op gezag van de
Rector Magnificus, dr. E. Sterken,
in het openbaar te verdedigen op
woensdag 7 maart 2012
om 16.15 uur

door
Carolin Curtze
geboren op 28 juni 1979
te Korbach, Duitsland

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Contents

You're walking.
And you don't always realize it,
but you're always falling.
With each step you fall forward slightly.
And then catch yourself from falling.
Over and over, you're falling.
And then catching yourself from falling.
And this is how you can be walking and falling
at the same time.

Laurie Anderson (1982). Walking and Falling. *Big Science*

Introduction | 1

Background

Locomotion is one of the most basic human motor activities. Over the course of time, from our first steps as a toddler to the end of early childhood, walking becomes a highly flexible and fully automated process. Even negotiating complex terrain becomes seemingly effortless. With the loss of a lower limb, this automated motor task is suddenly interrupted (Figure 1.1). After amputation, complex reorganization of postural (Geurts et al., 1991; 1992; Geurts and Mulder, 1994) and movement control (Hof et al., 2007) needs to take place.

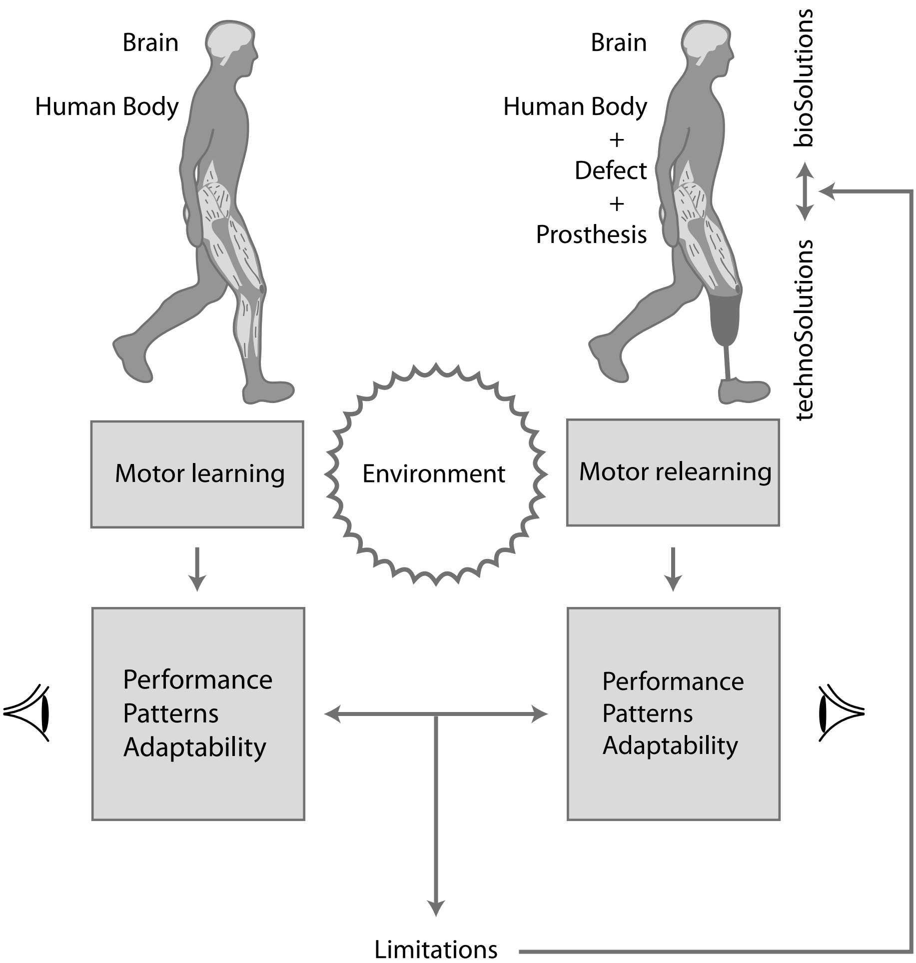

Figure 1.1 | Interaction between *brain, body, prosthesis* and *environment*.

Motor relearning occurs when a person with a lower limb amputation interacts with the environment; in this way she/he learns to handle the new dynamics of the motor system. In order to move effortlessly again, the person needs to learn to incorporate the properties of the altered body and the new limb into movement control. For instance, due to the loss of the limb the distribution of body weight is asymmetric, thereby changing the location of the center of mass in symmetric stance. The prosthetic limb, although it is lighter than the amputated section of the limb, will feel heavier to the patient, due to the lack of direct muscular control. Moreover, to produce movement, the prosthetic limb needs to be controlled indirectly; this forces the motor control system to develop effective new strategies to be able to function properly in everyday life (Vrieling et al., 2007; 2008a; 2008b; 2008c; 2008d; 2009).

Through comparative studies of the performance, movement patterns and adaptability of able-bodied controls and amputees, insights into the functioning and limitations of the complex *brain, body, prosthesis and environment* interaction are gained (Figure 1.1). By focussing on the limits of performance of lower limb amputees, crucial information on the nature of adaptation and its shortcomings can be obtained. Based on these insights, tailored training and rehabilitation programs can be developed (bioSolutions; Figure 1.1), as well as improvements of prosthetic design (technoSolutions; Figure 1.1).

In this thesis an integrative perspective of movement in lower limb amputees is presented, including aspects of *prosthetic design, environmental challenges and principles of motor control.*

Outline of the Thesis

The research presented in this thesis starts by characterizing prosthetic foot properties, based on a newly developed testing method (*Chapter 2*). An inverted pendulum-like apparatus is developed, for measurements of the mechanical properties of prosthetic feet and how these properties are affected by different types of shoes. Subsequently, the *patient–prosthesis* interplay is studied in *Chapter 3*. The coupling between the roll-over properties of a prosthetic device and the emergent (asymmetric) movement behavior is a central theme of this chapter.

The following *Chapters 4–7* deal with the problem of staying in dynamic balance when wearing a leg prosthesis. This series of studies provides insights into the adaptive motor behavior of amputees and healthy controls in complex motor tasks and challenging environments. In *Chapter 4*, a new measure for one-legged lateral balance control is introduced, in which balance is challenged by gradually reducing the lateral base of support. A specific feature of the test is that a broad variety of balance control mechanisms come into play. In *Chapter 5*, static balance of lower limb amputees is perturbed through external perturbations to their waist. By analyzing their balance response, the contribution of the prosthetic and sound limb to balance control is disentangled. In *Chapter 6*, the mechanisms of balance recovery from an evoked forward fall are studied. Preventing an impeding fall requires a rapid and spatially well-directed stepping movement, demanding a high degree of movement control. In *Chapter 7*, the dynamic walking stability of lower limb amputees is challenged by irregularities of the walking surface to elicit adjustment strategies.

Finally, in *Chapter 8*, the mental representation of body and movement of amputees are studied. An analysis of the constraints and plasticity of the body schema as response to peripheral changes was performed. By means of a laterality recognition task that is known to elicit implicit mental rotation of the subject's own body part, the effects of lower limb amputation and rotationplasty on mental rotation are studied.

In a concluding discussion in *Chapter 9*, the findings of the presented research are integrated and put into perspective.

Comparative Roll-over Analysis of Prosthetic Feet

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Carolin Curtze, At L. Hof, Helco G. van Keeken, Jan P. K. Halbertsma, Klaas Postema & Bert Otten
Journal of Biomechanics (2009) 42: 1746–1753

Abstract

A prosthetic foot is a key element of a prosthetic leg, literally forming the basis for a stable and efficient amputee gait. We determined the roll-over characteristics of a broad range of prosthetic feet and examined the effect of a variety of shoes on these characteristics. The body weight of a person acting on a prosthetic foot during roll-over was emulated by means of an inverted pendulum-like apparatus. Parameters measured were the effective radius of curvature, the forward travel of the center of pressure, and the instantaneous radius of curvature of the prosthetic feet. Finally, we discuss how these parameters relate to amputee gait.

Introduction

Given the design of a prosthetic foot, how does it match the function of a biological foot? In human plantigrade gait the foot rolls over the ground during each step, analogous to a wheel (Hansen et al., 2004a), while the stance leg acts like an inverted pendulum (e.g., Winter, 1995b; Winter, 1995a). The center of mass (CoM), located roughly at the level of the pelvis, travels in a series of arcs (Figure 2.1A). The overall motion can be described by a rocker-based inverted pendulum model, which, in contrast with a simple inverted pendulum with a peg, allows the center of pressure (CoP) to travel forward along the curved foot (Figure 2.1A–D). It has been shown that able-bodied people have curvatures of about 30% of the leg length during walking (McGeer, 1990; Hansen et al., 2004a). The energy cost of walking is lower on long, smoothly curved feet than on flat or pointy feet. It has been reported that the metabolic costs are optimal when walking on a foot with a curvature of 30% of the leg length (Adamczyk et al., 2006). Moreover, foot curvature has been demonstrated to be surprisingly invariant to walking speed, loads carried, and shoe heel height (Hansen et al., 2004a; Hansen and Childress, 2004; Hansen and Childress, 2005). On the other hand, considerable differences were found between gait initiation, steady-state walking, and gait termination (Miff et al., 2008). Here, the orientation of the curvature changed from "flexed" to "neutral" to "extended" (Figure 2.1C) through active muscle control.

The major challenge when designing prosthetic feet is to substitute the actions of the biological counterpart as efficiently as possible by means of a passive dynamic device. In this endeavor the design of prosthetic feet has become increasingly sophisticated in recent years. From this development arises an immediate need for a standard test method not only quantifying the mechanical properties of prosthetic feet, but further allowing a seamless translation of mechanical test data to performance data of gait experiments in individual patients.

A wide range of mechanical properties of prosthetic feet have been studied, such as stiffness (Van Jaarsveld et al., 1990; Lehmann et al., 1993), natural frequency (Lehmann et al., 1993), energy recovery and hysteresis (Van Jaarsveld et al., 1990; Postema et al., 1997; Geil et al., 2000), viscoelasticity (Geil, 2002), and material fatigue (Toh et al., 1993). While these studies have addressed im-

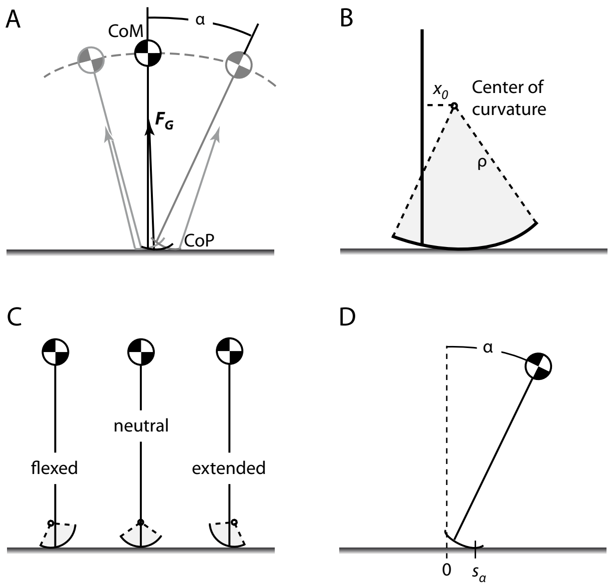

Figure 2.1 | Rocker-based inverted pendulum model. **(A)** Model of the leg as inverted pendulum supporting the center of mass (CoM). A ground reaction force (F_G) acts on the CoM, originating at the center of pressure (CoP). **(B)** Roll-over characteristics of a foot approximated by a constant radius of curvature (ρ), and fixed horizontal position of the center of the curvature (x_0). **(C)** The position of the rocker center (x_0) affects the orientation of the foot curvature. Able-bodied people actively manipulate the position of the effective rocker center by means of muscle control. A posterior shift of the center of curvature results in a more "flexed" orientation; this can be observed in able-bodied people during gait initiation. During steady-state walking the curvature maintains a "neutral" position. An "extended" orientation, i.e. an anterior shift of the center of curvature, is produced during the last step of gait termination. **(D)** General foot model with a variable curvature; $s(\alpha)$ is the location of the CoP on the ground as a function of shank angle α. The zero position is defined as the horizontal position of the shank at α=0. The instantaneous radius of curvature (s') is the first derivative of the forward travel (s) with respect to the shank angle (α).

portant properties of prosthetic feet, the resulting qualitative characteristics have often lacked a seamless translation into amputee gait characteristics due to separate measurement techniques. Furthermore, the nature of loading applied during testing is very different to that in walking. During testing the loading was increased at a constant ground angle, while during actual gait the ground angle changes during rollover. In a similar method, introduced by Hansen et al. (2000), the complex actions of the ankle-foot system are captured in an overall motion. Through quasi-static loading with a custom-made jig the authors estimated the roll-over shape, i.e. the effective curvature of prosthetic feet. The loading apparatus was modified in a later study (Hansen et al., 2004b); here a small weight was attached to the prosthetic foot and rolled over while a technician applied additional loading. This method has practical limitations with respect to the maximal load that can be applied. In addition, a rather short pendulum length was used so that the moments acting on the foot may be different to those during amputee gait; this might affect the effective roll-over shape. In this study we extend the work of Hansen et al. (2000; Hansen et al., 2004b) by introducing a novel inverted pendulum-like apparatus allowing continuous measurements. The technical requirements for the device testing method presented here are basically the same as those for a standard gait analysis, making intricate custom-made loading jigs and other costly solutions redundant.

The purpose of this study is to determine the roll-over characteristics of a range of prosthetic feet and to discuss their biomechanical implications for prosthetic gait. Furthermore, we will quantify the effect of different types of shoes on these characteristics. We expect the differences between the prosthetic foot models to be stronger than the effects imposed by the various shoes.

Methods

We designed an experiment in which we investigated the roll-over characteristics of a number of prosthetic feet in combination with different shoes by means of an inverted pendulum-like apparatus. Roll-over shapes of these foot–shoe combinations were simulated for a hypothetical subject with a body mass of 70 kg and a body height of 1.80 m.

Apparatus

The inverted pendulum-like apparatus consisted of a shaft, with a prosthetic foot attached to the lower end and a mass (m) of 70 kg mounted to the upper end of the shaft (Figure 2.2). The pendulum length (l), the distance between the foot sole and the CoM of the added weight, was 0.98 m, a typical leg length for a person of 1.80 m body height. A custom-made rig provided lateral guidance during testing with a minimum of friction; further it restricted the anterior–posterior range to predefined limits.

Seven prosthetic feet of three different manufacturers were included in this study (Table 2.1): Endolite (Esprit, Navigator), Össur (Flexfoot, Vari-Flex), and Otto Bock (1C40, 1D10, 1D35). All feet were right sided and sized 270 mm. Each prosthetic foot was tested under 4 different conditions: (1) without shoe, (2) with a men's leather shoe, (3) with a running shoe, and (4) with a hiking boot.

The prosthetic feet were mounted in neutral alignment, i.e. the feet were flat on the ground when the shaft was vertical under zero-load. The alignment was adjusted for shoe heel height to obtain a vertical standing position of the shaft under zero-load.

The ground reaction force ($\boldsymbol{F_G}$) and the position of the center of pressure were measured with an AMTI force plate, sampled at 1000 Hz. Two reflective markers were placed on the shaft. The markers were tracked by an eight-camera VICON motion analysis system at a sampling frequency of 100 Hz. The data was then further processed with Matlab. This delivers, amongst others, shank angle (α) and ankle moment data.

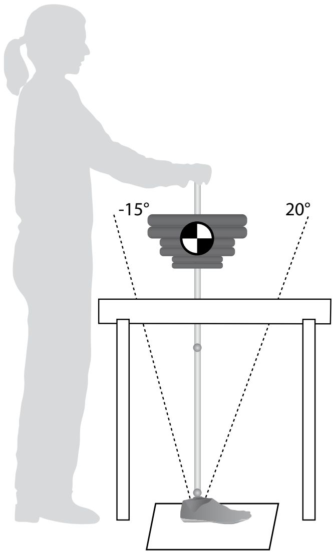

Figure 2.2 | The inverted pendulum-like apparatus. The apparatus consisted of a prosthetic foot, a shaft, and a 70-kg mass. The CoM of the weight was mounted at a height of 0.98 m. Two reflective markers were placed on the shaft to determine the shank angle. The ground reaction force ($\boldsymbol{F_G}$) and the position of the CoP were measured with a force plate. Roll-over characteristics were determined over a range of −15° to 20°. A custom-made rig provided lateral guidance during roll-over.

Table 2.1 | Prosthetic feet and shoes used for testing.

Manufacturer	Model		Material & Classification
Endolite	Esprit		Carbon fiber Mobility Grades 1–4
	Navigator		Snubber, ankle ball PTFE (tetrafluoroethylene)-coated keel Mobility Grades 2 and 3
Össur	Talux		Carbon fiber Tarsal core Mobility Grades 2 and 3
	Vari-Flex		Carbon fiber Mobility Grades 1–4
Otto Bock	1C40/ C-Walk		Carbon fiber Mobility Grades 3 and 4
	1D10/ Dynamic foot		Contoured core and foam Mobility Grades 1 and 2
	1D35/ Dynamic motion		Plastic spring and foam Mobility Grades 2 and 3
Nike	Air		Running shoe
Vertice			Leather shoe
Scarpa	ZG40		Hiking boot

Experimental procedure

The experimenter applies a horizontal force to the top weight necessary for an angular velocity of ±10 deg/s, making the foot roll over from heel to toe and back. The measurement range was −15° to 20° with respect to the absolute vertical, corresponding to heel-contact and toe-off. The testing procedure was repeated three times for each foot–shoe combinations. (A video of the experimental procedure is available as Supplementary Material.)

Data analysis

Effective radius of curvature (ρ)

During part of the stance phase a biological ankle-foot system acts like a smoothly curved solid object. The CoP progresses forward from heel to toe, similar to that of a rolling wheel with a particular radius. Hansen et al. (2004a) proposed a method that allows estimating the effective "ankle-foot roll-over shape" of an ankle-foot system from CoP data. The strength of this method is its universal applicability, ranging from a simple rolling wheel, to deformable objects and complex multi-joint systems. By transforming successive CoP location data from a laboratory-based coordinate system into a shank-based coordinate system the effective curvature geometry can be determined. The resulting ankle-foot roll-over shape is similar to a circle and reflects the overall motion of the system. The radius of curvature is determined by fitting the best-fit circular arc to the transformed data (Hansen et al., 2004a).

Center of curvature (x_0)

The horizontal position of the center of curvature (x_0) with respect to the shank determines the orientation of the foot curvature (Figure 2.1B and C). For feet with an extended orientation the center of curvature is anterior to the shank, while for feet with a flexed orientation it is posterior.

Forward travel (s)

As the foot is rolled over, the CoP travels along the curvature of the foot (Figure 2.1D). A fast forward travel (s) as a function of shank angle (α) corresponds to a very flat/large radius of curvature, while a highly curved/small radius of curvature produces only little forward travel as a function of shank angle. A large radius of curvature is to be considered as more stable than a small radius of curvature.

The instantaneous forward travel (s) was determined for shank angle (α) ranging between −15° and 20°. In theory, for feet with a perfectly constant curvature

$$s = \alpha\rho + x_0$$

Instantaneous radius of curvature (s')

The instantaneous radius of curvature (s') is determined as the first derivative of the forward travel (s) on the ground with respect to the shank angle (α)

$$s' = \frac{ds}{d\alpha}$$

It is related to the position-dependent stability of the foot, which increases with the instantaneous radius of curvature (s'). In contrast to the effective radius of curvature (ρ), the instantaneous radius of curvature (s') refers only to the CoP displacement over the floor.

Comparative single case study

In addition to the mechanical testing of the different foot–shoe combinations, we performed a single case study to get an estimate for the agreement between roll-over characteristics measured in an actual amputee and those obtained through mechanical testing.

The transfemoral amputee had a body weight of 68 kg and a body height of 1.82 m. The subject was fitted with a four-bar linkage prosthetic knee (Proteval), an Otto Bock 1C40 foot (size 27 cm), and a leather shoe. Reflective markers were placed in positions corresponding to those used during mechanical testing. The patient was asked to walk at a comfortable walking speed.

Results

Exemplary roll-over shapes superimposed on contour drawings of prosthetic feet are shown in Figure 2.3A. The depicted roll-over shapes of the Esprit foot (left) and the Vari-Flex foot (right) appear to be geometrically distinct. While the roll-over shape of the Esprit foot is characterized by a rather flat middle section, the roll-over shape of the Vari-Flex foot is more circular. Further differences become evident when studying the formation of roll-over shapes with respect to the angular displacement. For the Esprit foot the CoP at zero-degree shank angle (circle) is located more proximal to the ankle than that of the Vari-Flex foot. Further, the clustering of stars (five-degree increments) at the tails of the left roll-over shape indicates that during roll-over the CoP remains at the heel, and afterwards quickly travels forward. In the other example, by contrast, the stars are more evenly distributed, indicating a rather constant curvature. These differences become even more prominent when examining the forward travel (Figure 2.3B). The forward travel of the Esprit foot is characterized by an S-shaped progression, with little forward travel in the two plateau phases. The more forward travel there is, i.e., the steeper the slope, the larger is the radius of curvature. The forward travel changes during roll-over, it increases in the middle section, thereby making the foot more stable at this shank angle. The almost linear forward travel of the Vari-Flex foot suggests a nearly constant radius of curvature (Figure 2.3C). Here, the instantaneous radius of curvature (s'), i.e. the slope of the forward travel, deviates little from the effective radius of curvature (ρ, stippled line). The S-shaped progression of forward travel in the Esprit foot is reflected in the three-phased structure of the instantaneous radius of curvature.

The effective radius of curvature (ρ, Figure 2.4A) was found to differ widely between the tested foot models. Shoes cause a reduction in radius of curvature. Interestingly, feet with a small radius of curvature appeared almost invariant to the effect of applying shoes. Along with the effective radius of curvature (ρ), the roll-over characteristics of a foot are determined by the horizontal position of the center of curvature (Figure 2.4B). The center of curvature (x_0) varied widely among foot models. It shifted closer towards the shaft as response to the different shoes. As a result the roll-over shapes created by the foot–shoe combinations have a rather “neutral” orientation.

Figure 2.5A illustrates the effect of shoes on the forward travel (*s*) of the Esprit foot. The characteristic S-shaped forward travel is straightened by means of different shoes. The hiking boot appeared to modulate the forward travel the

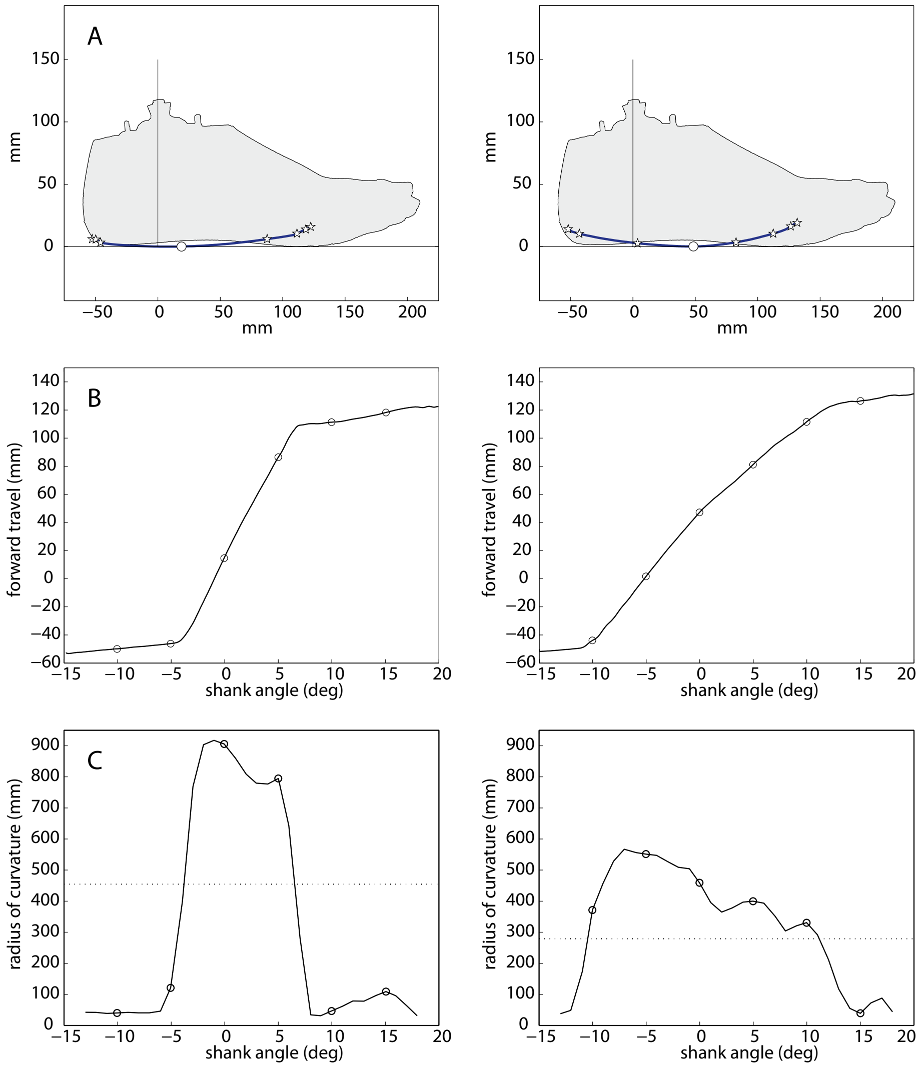

Figure 2.3 | Roll-over characteristics of prosthetic feet. **(A)** Exemplary roll-over shapes (solid line) of the no-shoe condition of the Esprit foot (left) and the Vari-Flex foot (right). The circles mark the instantaneous position of the CoP in a shank-based coordinate system at zero-degree shank angle; stars denote the position of the CoP in five-degree increments. **(B)** The forward travel (*s*) of the CoP during roll-over. **(C)** The instantaneous radius of curvature (*s'*) is determined as the derivative of the forward travel (*s*). The stippled line indicates the effective radius of curvature (ρ). The larger the radius of curvature, the more stable is the foot.

strongest. The instantaneous radius of curvature (*s'*) shows these differences between foot models by a reduction in range (Figure 2.5B).

The general effects of shoe models on the forward travel (*s*) are given in Figure 2.6A. Differences between foot models were prominent (Figure 2.6B); while some feet are characterized by a distinct S-shape, others have a rather linear forward travel.

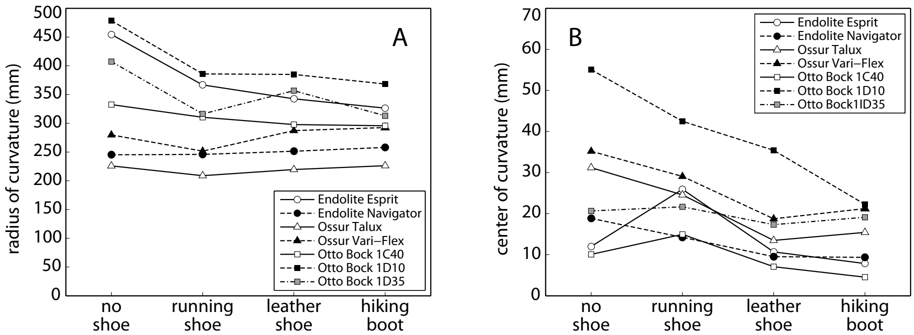

Figure 2.4 | Effective radius of curvature (ρ) and center of curvature (x_0). **(A)** The radius of curvature was estimated through fitting a circular arc to shank-based transformed CoP data. The radius of curvature is an indicator for the overall stability of feet; for stable feet the radii are larger than for unstable feet. The effective radius of curvature is generally smaller with each of the shoes compared to the no-shoe condition. **(B)** The horizontal position of the center of curvature determines the orientation of the roll-over shape. The more forward the position of the center of curvature (large x_0), the more extended the foot. With one exception, for the different shoes the center shifts towards the shank, compared to the no-shoe condition, resulting in a more neutral orientation of the roll-over shape.

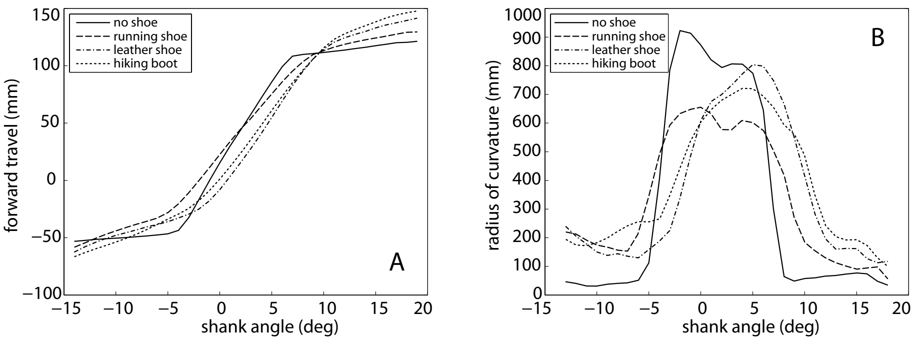

Figure 2.5 | Forward travel (*s*) and instantaneous radius of curvature (s') of the Esprit foot. **(A)** Shoes modulated the forward travel towards a more linear progression. **(B)** The range of the instantaneous radius of curvature is decreased when wearing shoes, compared to the no-shoe condition, resulting in a more constant radius of curvature.

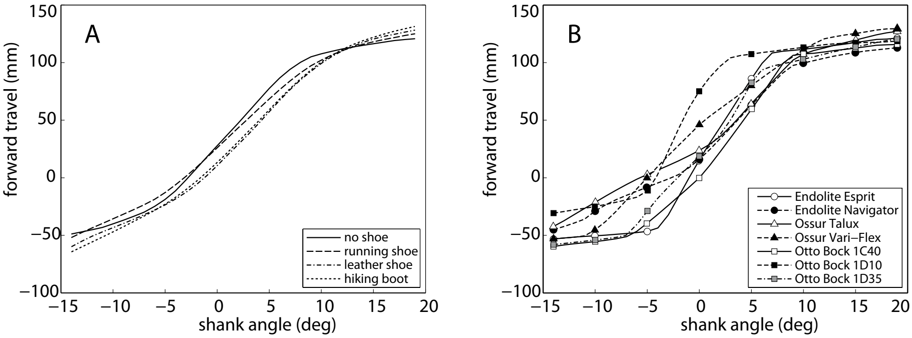

Figure 2.6 | Forward travel (*s*). **(A)** The effects of shoe on the forward travel of the CoP; data presented are the overall means for all foot models. **(B)** The forward travel of different foot models measured without shoe.

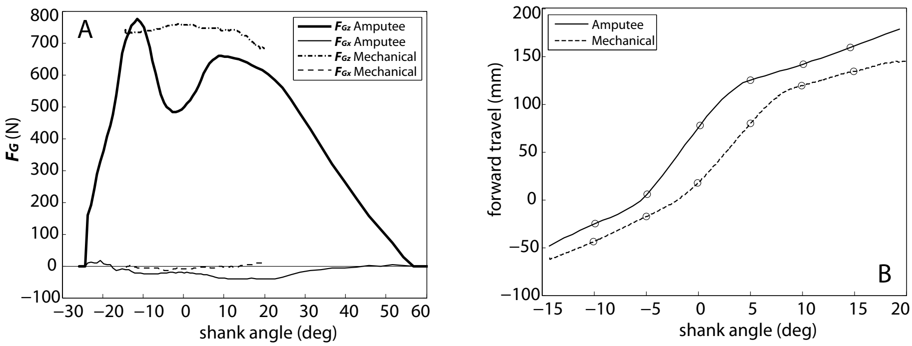

Figure 2.7 | Comparative illustration of $\boldsymbol{F_G}$ and forward travel. **(A)** The vertical ($\boldsymbol{F_{Gz}}$) and the forward ($\boldsymbol{F_{Gx}}$) component of the $\boldsymbol{F_G}$ of the amputee (prosthetic side) and during mechanical testing. **(B)** The forward travel (*s*) of the CoP during roll-over.

In amputee gait the prosthetic knee is in full extension during roll-over, making the inverted pendulum apparatus a relevant model of the shank angle of the amputee. The ground reaction forces acting on the prosthetic foot during mechanical testing versus amputee gait are shown in Figure 2.7A. The vertical ground reaction force ($\boldsymbol{F_{Gz}}$) of amputee gait is characterized by a two-peak shape, corresponding to heel-strike and toe-off. The measurement range of mechanical testing (–15° to 20°) covers the angular area corresponding to those two peaks. However, the vertical forces produced during mechanical testing are more even and lack the heel-strike/toe-off peaks. Nonetheless, contrasting the forward travel (*s*) obtained by mechanical testing with that of a transtibial amputee revealed strong similarities (Figure 2.7B). The gait measurements revealed an effective radius of curvature (ρ) of 352 mm, compared to 322 mm found in mechanical testing.

Discussion

Mechanical testing of a broad range of prosthetic feet revealed that the typical radius of curvature varied around 312 mm (Figure 2.4A). This effective radius of curvature agrees closely with that found in able-bodied people (Hansen et al., 2004a). In an experiment on the effect of different rocker-foot curvatures on the metabolic costs of walking of able-bodied subjects, a curvature of about 0.3*l* (leg length) was shown to be energetically advantageous (Adamczyk et al., 2006). The shoes induced a small reduction in radius of curvature, which made the feet more similar to each other. This reduction in curvature will probably be related to an interaction between shoe stiffness and the unloaded shape of the sole of these shoes. Hansen et al. (2000) reported that shoes did not have as significant effect on the curvature of a roll-over shape, but offset the roll-over shape by the sole thickness. However, Hansen et al. (2000) used just a single shoe model in their experiment, a soft-soled Nike ACG hiking shoe. The modulating effects of different shoe models on the forward travel, as presented here, indicate that different shoes indeed lead to change in curvature and not just an offset of roll-over shape.

The horizontal position of the center of curvature, which reflects the orientation of the foot curvature, differs widely between feet. Shoes generally reduced the extended orientation of the foot curvature, which is due to heel height of the shoes. The resulting, almost neutral orientation agrees with values found in able-bodied people during steady-state walking (Miff et al., 2008). It is to be noted that the position of the center of curvature of a prosthetic foot can be influenced through the alignment of the prosthesis. Mounting a prosthetic foot in a more dorsally flexed orientation under zero-load should be beneficial for gait initiation, while a more extended orientation serves gait termination (Hansen et al., 2004a; Miff et al., 2008). For this reason, each prosthetic foot is to be aligned according to its inherent properties and the needs of the patient. This sensitivity of prosthetic foot properties to alignment may also explain why numerous studies failed to identify consistent differences in amputee gait, despite the mechanical differences between the tested prosthetic feet (e.g. Zmitrewicz et al., 2006; Postema et al., 1997; Powers et al., 1994).

However, modeling the roll-over shape as a circle with a uniform radius of curvature and fixed center neglects that feet have no constant curvature. We showed that the forward travel does not progress linearly as it would be the case in feet with a constant curvature. In fact, the forward travel follows an S-shaped curve, in which three sections can be discriminated: a steeper middle section (−5° to 10°), and flattened begin and end sections. The increase in forward travel in the middle section implies a flat/large instantaneous radius of curvature. This suggests a greater stability in stance, which can be assumed to be beneficial for standing stability. Besides, feet with a large travel give a reduced initial peak of vertical ground reaction forces for the contralateral limb (Hansen et al., 2006; Adamczyk et al., 2006). This may eventually be explained by an increased gain in velocity of the CoM during roll-over and an enlarged directional change of the CoM for feet with a small travel. We propose forward travel as another important measure of prosthetic foot characteristics, due to its implications for gait and stability. We found that forward travel was strongly distinct between prosthetic feet. Shoes imposed slight effects on the forward travel—they made it more constant. This is directly related to the fact that shoes have also curvatures when not loaded.

We found that the available prosthetic feet have widely different biomechanical properties, including strongly distinct roll-over shapes. Shoes modulate these roll-over shapes slightly, but most likely functionally significantly. Since predicting the effect of shoes on prosthetic feet is not straightforward, measurements are indispensable. Here, our study offers a simple method to determine roll-over properties of user-defined foot–shoe combinations. The required equipment is that of a standard gait lab, comprising a force plate and a motion tracking system. In this study the inverted pendulum, which simulated the prosthetic leg, was moved in a strictly sagittal plane with the prosthetic foot pointing straight forward. In clinical practice it is sometimes necessary to mount the prosthetic foot in a deviating orientation. The effects of different outlines of a prosthesis on the roll-over characteristics can be simulated exactly in the same way.

While this method allows emulating the body weight of a person that acts on a prosthetic foot during roll-over, it cannot accurately reproduce the complex multi-joint dynamics that occur during amputee gait. Despite the differences in vertical ground reaction forces, the forward travel curves for amputee gait and mechanical testing were remarkably parallel. The offset between both curves can be explained by differences in prosthesis alignment. With respect to the good agreement of measured roll-over characteristics, the proposed method represents a valuable means to objectively compare foot–shoe combinations. Mechanical testing has advantages compared to measurements in amputees; most importantly, it is not affected by step-to-step variability in amputees. Furthermore, the patient might adjust his gait pattern to the roll-over characteristics of a foot–shoe combination, which would blur the actual differences in roll-over.

In the future it would be interesting to systematically manipulate single properties of prosthetic feet and test their effect on amputee gait. This would give us valuable information on the dynamics of amputee gait and prosthetic design.

Suppliers

- Advanced Mechanical Technology Inc., 176 Waltham Street, Watertown, MA 02472-4800, USA
- Vicon Motion System, 14 Minus Business Park, West Way, Oxford OX2OJB, UK
- The MathWorks Inc., Crystal Glen Office Centre, 39555 Orchard Hill Place, Suite 280, Novi, MI 48375, USA

Conflict of interest

No financial support was received from the prosthetic industry. Prosthetic feet were provided on a loan basis.

Acknowledgements

We gratefully acknowledge Henk Zijlstra for his expertise in prosthetics, Ronald Davidsz for his technical assistance, Bert Slootmaker for the construction of the apparatus, and Cojanne Kars for her assistance with data collection.

Further, we thank Bauerfeind AG, Össur hf, and Otto Bock HealthCare GmbH for providing the prosthetic feet. Reprint of the images in Table 2.1 with kind permission of Otto Bock HealthCare GmbH and Chas A Blatchford & Sons Ltd. is acknowledged.

Supporting Information

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.jbiomech.2009.04.009.

Determining Asymmetry of Roll-over Shapes in Prosthetic Walking

3

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Journal of Rehabilitation Research & Development (2011) 48: 1249–1260

Abstract

How does the inherent asymmetry of the locomotor system in people with lower-limb amputation affect the ankle-foot roll-over shape of prosthetic walking? In a single-case design, we evaluated the walking patterns of six people with lower-limb amputation (3 transtibial and 3 transfemoral) and 3 matched able-bodied controls. We analyzed the walking patterns in terms of roll-over characteristics and spatial and temporal factors. We determined the level of asymmetry by roll-over shape comparison (root-mean-square distance) as well as differences in radius of curvature. In addition, we calculated ratios to determine spatial and temporal asymmetries and described different aspects of asymmetry of roll-over shapes. All participants showed some level of asymmetry in roll-over shape, even the able-bodied controls. Furthermore, we found good intralimb reproducibility for the group as a whole. With respect to spatial and temporal factors, the participants with transtibial amputation had a quite symmetrical gait pattern, while the gait in the participants with transfemoral amputation was more asymmetrical. The individual ankle-foot roll-over shapes provide additional insight into the marked individual adjustments occurring during the stance phase of the sound limb. The two methods we present are a suitable measure for determining asymmetry of roll-over shapes; both methods should be used complementarily.

Introduction

While as a rule, unimpaired walking is 1 reasonably symmetric, walking with a prosthetic limb is characterized by a marked asymmetry of limb movements (Hof et al., 2007). In people with lower-limb amputation, the locomotor system is inherently asymmetric in its construction with a passive prosthetic and an active, muscle-controlled sound limb. This results in multiple adjustments in gait pattern on the level of temporal dynamics (e.g., in people with transtibial amputation). The stance phase on the sound limb is prolonged, while on the spatial level the lateral stability margin of stepping is increased on the side of the prosthetic limb by placing the foot farther outside, thereby making walking with a prosthesis more stable (Hof et al., 2007). In addition to these spatio-temporal adjustments, changes in interlimb coordination (Donker and Beek, 2002) as well as alterations in joint kinetics have been reported (Beyaert et al., 2008; Silverman et al., 2008).

In the stance phase of walking, the foot rolls over the ground from heel to toe, analogous to a rolling wheel. Hansen et al. (2004a) described a method for capturing the overall motion of the ankle-foot system in so-called "effective ankle-foot roll-over shapes," which form the basis of the rocker-based inverted pendulum model of walking. Ankle-foot roll-over shapes are determined by transforming the successive center of pressure (CoP) displacement from a laboratory-based coordinate system into a shank-based coordinate system. A remarkable invariance of these nearly circular roll-over shapes has been reported for able-bodied subjects during walking. The literature shows that the preferred ankle-foot curvature is unaffected by shoe-heel height (Hansen and Childress, 2004), loads carried (Hansen and Childress, 2005), and different walking speeds (Hansen et al., 2004a). Adjustments are found in the orientation of knee-ankle-foot roll-over shapes only when starting and stopping (Miff et al., 2008).

This apparent invariance of roll-over shapes, as found in able-bodied subjects, has been suggested as an "ideal" design feature for prosthetic feet (Hansen et al., 2003). Hansen et al. (2003) show that transtibial prostheses are aligned (by a prosthetist) to match as closely as possible this ideal roll-over shape. Hansen et al. (2003; Hansen et al., 2000) have also suggested that prosthetic feet should mimic the roll-over shapes of a natural foot, which they appear to do with respect to the radius of curvature when mechanically tested (*Chapter 2*; Curtze et al., 2009). Hence, prosthetic feet mimicking the natural roll-over shape of a nondisabled ankle-foot system should result in a natural roll-over shape under the sound limb. So far, the literature does not describe how the roll-over shape of the prosthetic limb affects the roll-over shape of the sound limb in people with lower-limb amputation. To this end, we assessed the individual differences in roll-over shapes of 3 people with transtibial and 1 transfemoral amputation during steady-state walking. We present two methods to quantify differences in roll-over shapes: (1) based on the radius of curvature and (2) based on the quantitative coordinate comparison of the curve structures, i.e., the average distance between curves. Based on these measures, we determined the interlimb asymmetry and the intralimb reproducibility of roll-over shapes.

For a general characterization of each participant's walking patterns, we report temporal gait parameters because reference values for these characteristics are well established in the literature for all three populations (van der Linden et al., 1999; Vanicek et al., 2009a; Lamoth et al., 2010).

Methods

Participants

Six participants with amputation (three transtibial and three transfemoral) participated in the single-case design study. All participants with amputation were experienced and able walkers. They performed the test with the prostheses that they use on a daily basis. Three nondisabled participants formed the control group. All nine participants were male; Table 3.1 gives further participant characteristics.

Apparatus

We measured the ground reaction forces during a complete stride cycle using two force plates (AMTI; Watertown, Massachusetts) sampling at a rate of 100 Hz. We placed reflective markers on the lateral epicondyle of the knee and the lateral malleolus of both limbs. On the prosthetic limb, we placed the mark-

Table 3.1 | Participant data.

	Height (m)	Weight (kg)	Age (yr)	Time Since Amputation (yr)	Side of Amputation	Cause of Amputation	Prosthesis Components & Shoes
Controls							
1	1.78	78	41	-	-	-	Leather shoe
2	1.88	88	56	-	-	-	Leather shoe
3	1.86	82	56	-	-	-	Leather shoe
Transtibial amputees							
4	1.82	86	64	2	right	vascular	C-Walk (Otto Bock) Leather shoe
5	1.81	83	67	8	left	vascular	C-Walk (Otto Bock) Leather shoe
6	1.87	89	60	3	right	vascular	1D35 (Otto Bock) Leather shoe
Transfemoral amputees							
7	1.77	88	51	9	left	vascular	3R106=ST (Otto Bock) Multiflex (Endolite) Running shoe
8	1.82	68	43	26	right	cancer	Acphapend (Proteval) C-Walk (Otto Bock) Running shoe
9	1.87	90	37	12	left	trauma	TGK-4P10 Graph-Lite (T-lin) C-Walk (Otto Bock) Trekking shoe

ers at the corresponding positions, allowing determination of the shank angle. An eight-camera motion capture system (Vicon; Oxford, United Kingdom) tracked the reflection markers at a sampling rate of 100 Hz. We determined the temporal events of the gait cycle (heel-contact and toe-off) using foot switches (AURION S.r.l.; Milano, Italy).

Experimental procedure

We asked the participants to walk at their self-selected comfortable walking speed. We needed three trials per participant with "clean" hits (i.e., full foot support) on both force plates.

Data analysis

We used a Woltring filter with a predicted MSE (mean square error) value of 10 to filter the marker trajectories (Woltring, 1985). We further processed the data using MATLAB (MathWorks; Natick, Massachusetts).

Ankle-foot roll-over shape

We analyzed three stride cycles for each participant. We analyzed roll-over shapes in two dimensions, reflecting the overall walking direction. By transforming the successive CoP data (from heel-contact to opposite heel-contact) from a laboratory-based into a shank-based coordinate system (Figure 3.1A and B), we determined the effective roll-over shapes. We set the threshold for heel-contact to one-third of body weight. We estimated the effective radius of curvature by fitting a best-fit circular arc to the transformed position data (*Chapter 2*; Curtze et al., 2009; Hansen et al., 2004a). As the radius of curvature is a geometrical entity, we determined it by averaging over shape, thus giving equidistant points equal weight. We defined the origin of the shank-based coordinated system as the intersection of the extension of the shank and the floor at vertical shank angle.

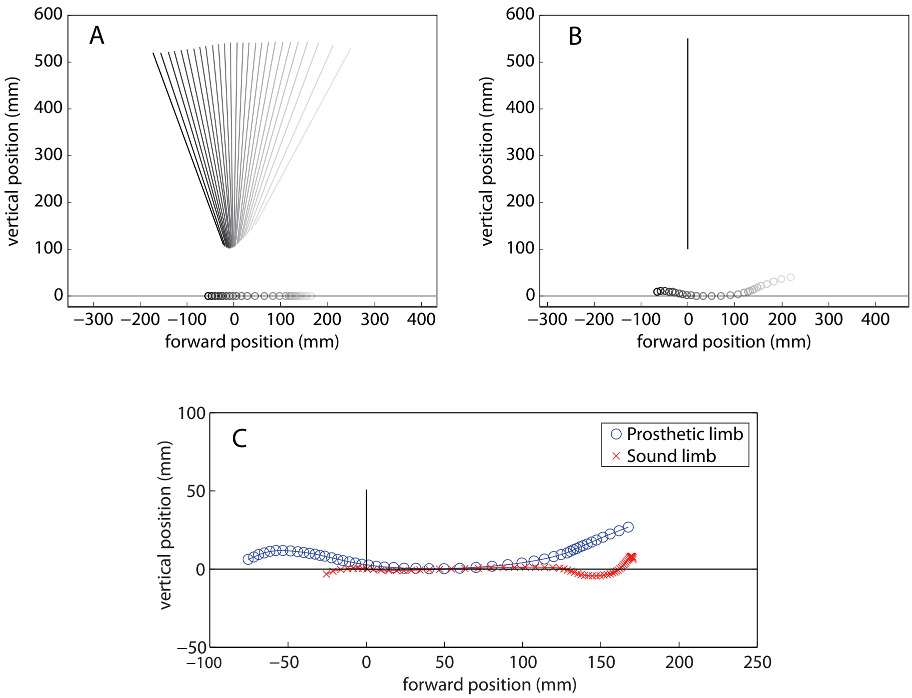

Figure 3.1 | Transformation of successive center of pressure data from **(A)** laboratory-based coordinate system into **(B)** shank-based coordinate system. **(C)** Resampled shank-based roll-over shapes to calculate r.m.s. distance between curves. Presented are roll-over shapes of sound and prosthetic limbs of participant with transfemoral amputation (participant 8).

Comparison of ankle-foot roll-over shapes

We calculated the agreement between ankle-foot roll-over shapes as the average distance of two curves. The curves were resampled based on the distance travelled along the curve with a step size of 1 mm. Starting at the origin, the curves are resampled in anterior and posterior direction. As each roll-over curve may have a different overall length, we compared the resampled curves over their shared length by determining the root-mean-square (r.m.s.) distance in millimeters (Figure 3.1C).

We calculated the intralimb reproducibility, as well as the interlimb asymmetry, by comparing all three roll-over shapes of a limb with each other; then we determined the mean of the average r.m.s distance. Similarly, we calculated the interlimb asymmetry by comparing all 3 × 2 roll-over shapes of the left and right or sound and prosthetic limbs, resulting in 32 comparisons; finally, we determined the mean of the average r.m.s. distance. The r.m.s. distance is 0 for identical curves, and its value increases as the two curves become more different.

Spatial and temporal gait characteristics

We calculated gait velocity and stride time over a complete stride cycle. Calculated spatial and temporal parameters included step length, stance time (from heel-contact to toe-off), and double-support time (from heel-contact to opposite toe-off) for each limb. We quantified temporal gait symmetry by calculating the ratio between the mean stance time of both limbs as well as the ratio between the mean double-support time of both limbs. A ratio of 1 indicates perfect temporal symmetry.

Results

The temporal asymmetry in walking appeared to increase with the level of amputation (Table 3.2). While each of the controls and the participants with transtibial amputation showed a relatively symmetric stance time ratio, the stance time on the sound limb appeared to be prolonged in the participants with transfemoral amputation. Accordingly, the participants with transfemoral amputation showed a prolongation in double-support time on the prosthetic limb, while we observed no marked differences in double-support time for the participants with transtibial amputation (except for one, participant 6) or for the controls.

Figures 3.2–3.4 illustrate the roll-over shapes of each participant; three trials superimposed onto each other. Good intralimb reproducibility is indicated by the almost identical roll-over shapes; the three trials per limb nearly perfectly line up with each other, indicating that the overall motion of the ankle-foot system was almost identical. This good agreement is confirmed by the low r.m.s. distance between the curves (Table 3.3). Comparison of the roll-over shapes between limbs, however, revealed marked differences (Figures 3.2–3.4). The r.m.s. distances found for the between-limb comparison were larger than those found for the intralimb reproducibility (Table 3.3).

Table 3.2 | Gait characteristics.

	Velocity (m/s)	Stride time (s)	Step length (m)			Stance time (% of stride)			Double support time (% of stride)		
			Sound/ Left	Prosthetic/ Right	Ratio	Sound/ Left	Prosthetic/ Right	Ratio	Sound/ Left	Prosthetic/ Right	Ratio
Controls											
1	1.52 (0.01)	1.02 (0.01)	0.80 (0.01)	0.76 (0.02)	1.05	60.72 (1.08)	63.64 (0.99)	0.95	12.02 (0.64)	12.34 (1.20)	0.98
2	1.21 (0.004)	1.25 (0.02)	0.79 (0.03)	0.72 (0.02)	1.10	62.51 (1.23)	61.96 (0.89)	1.01	11.96 (0.61)	12.50 (0.56)	0.96
3	1.38 (0.004)	1.16 (0.01)	0.82 (0.02)	0.78 (0.02)	1.05	61.89 (0.45)	63.90 (0.58)	0.97	12.89 (0.13)	12.89 (0.86)	1.00
Transtibial amputees											
4	1.06 (0.05)	1.26 (0.04)	0.74 (0.13)	0.66 (0.01)	1.12	63.57 (1.11)	63.02 (2.66)	0.99	13.44 (1.22)	13.15 (2.43)	0.97
5	1.24 (0.03)	1.13 (0.03)	0.76 (0.13)	0.64 (0.15)	1.19	66.37 (0.29)	62.83 (1.47)	0.95	14.76 (0.71)	14.44 (1.07)	0.98
6	1.21 (0.06)	1.24 (0.04)	0.73 (0.03)	0.77 (0.01)	0.95	63.47 (1.13)	61.03 (0.45)	0.96	11.03 (0.45)	13.47 (1.13)	1.22
Transfemoral amputees											
7	1.00 (0.08)	1.26 (0.03)	0.61 (0.05)	0.89 (0.20)	0.69	67.72 (1.93)	58.43 (2.50)	0.86	11.89 (1.35)	14.27 (2.70)	1.21
8	1.22 (0.02)	1.14 (0.02)	0.70 (0.01)	0.69 (0.01)	1.01	66.38 (1.52)	55.25 (1.49)	0.83	9.64 (1.46)	11.99 (0.45)	1.27
9	1.21 (0.05)	1.23 (0.04)	0.68 (0.05)	0.80 (0.05)	0.85	64.77 (7.03)	60.00 (6.03)	0.94	11.75 (3.24)	13.03 (1.99)	1.19

mean (S.D.)

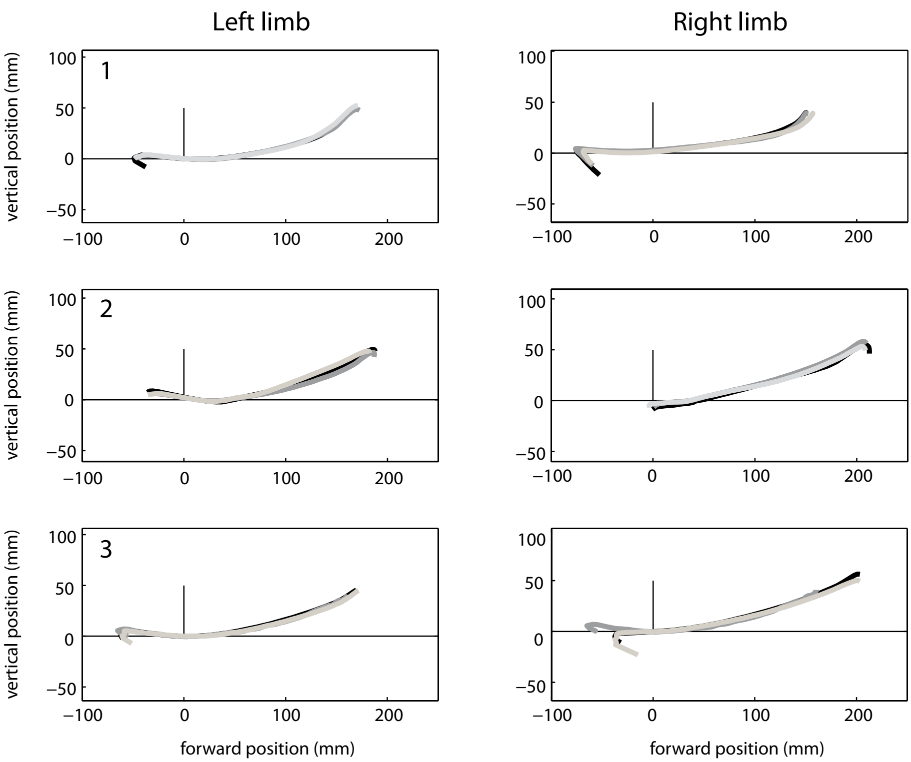

Figure 3.2 | Roll-over shapes of three trials for left and right limb in able-bodied controls (participants 1–3).

Interlimb comparison of the radius of curvature revealed marked asymmetries for most of the participants (Table 3.3). Only two of the participants with transtibial amputation (participants 4 and 6) showed symmetry with respect to radius of curvature (0.97 and 1.11, respectively). Visual inspection revealed that some roll-over shapes showed strong deviations from a circular segment (participants 8 and 9). Consequently, estimates of the radius of curvature vary considerably (Table 3.3). The differences were most prominent in participants 8 and 9, who had a high self-selected walking speed (Table 3.2). In order to walk fast, they pushed themselves up in the late stance phase of the sound limb, thereby providing additional ground clearance for the swinging prosthetic limb. These adjustments in gait pattern are known as "vaulting." Through excessive ankle plantar flexion, the patient gains height while the ro-

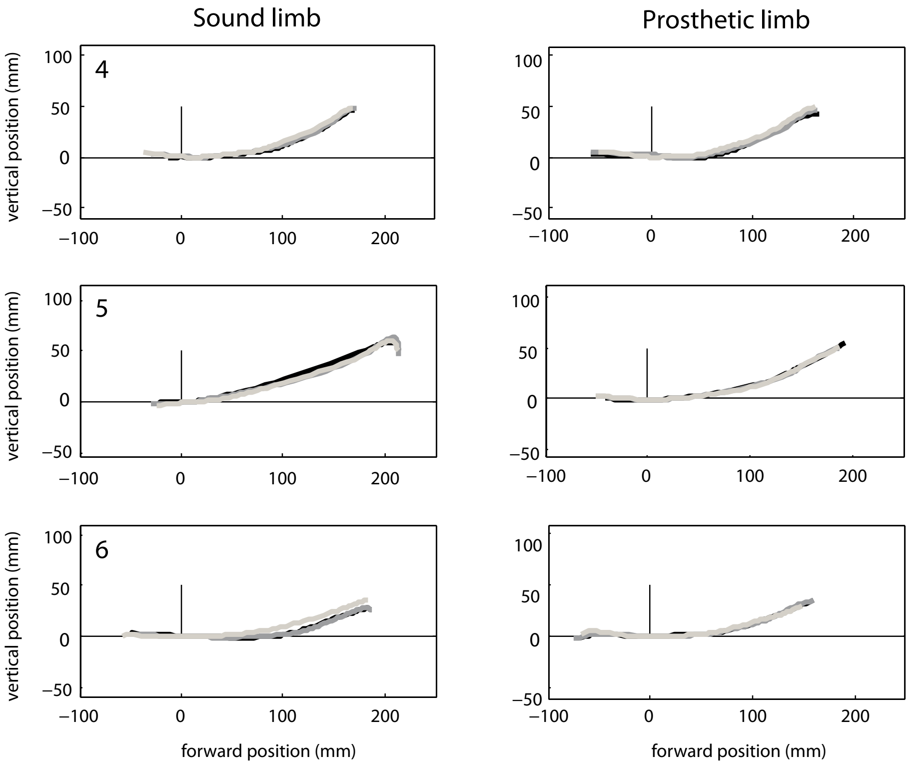

Figure 3.3 | Roll-over shapes of three trials for sound and prosthetic limbs in participants with transtibial amputation (participants 4–6).

tation of the shank angle is slowed and the CoP travels forward. This movement results in a flat roll-over shape, 1 with a very large radius (low curvature). Furthermore, video analyses revealed strongly asymmetric arm movement in participant 9. In the late stance phase on the sound limb, this participant pushed himself up and supported this motion by swinging the ipsilateral arm high.

Two of the participants with transfemoral amputation (participants 8 and 9) walked with the same type of prosthetic foot (Table 3.1). Interestingly, the resulting roll-over shapes of the prosthetic limb still differed substantially in form. This could be caused by differences in prosthesis outline and shoe model, as well as by differences in multijoint dynamics in combination with body weight differences. The "hooks" at the beginning (participants 1 and 3) or the

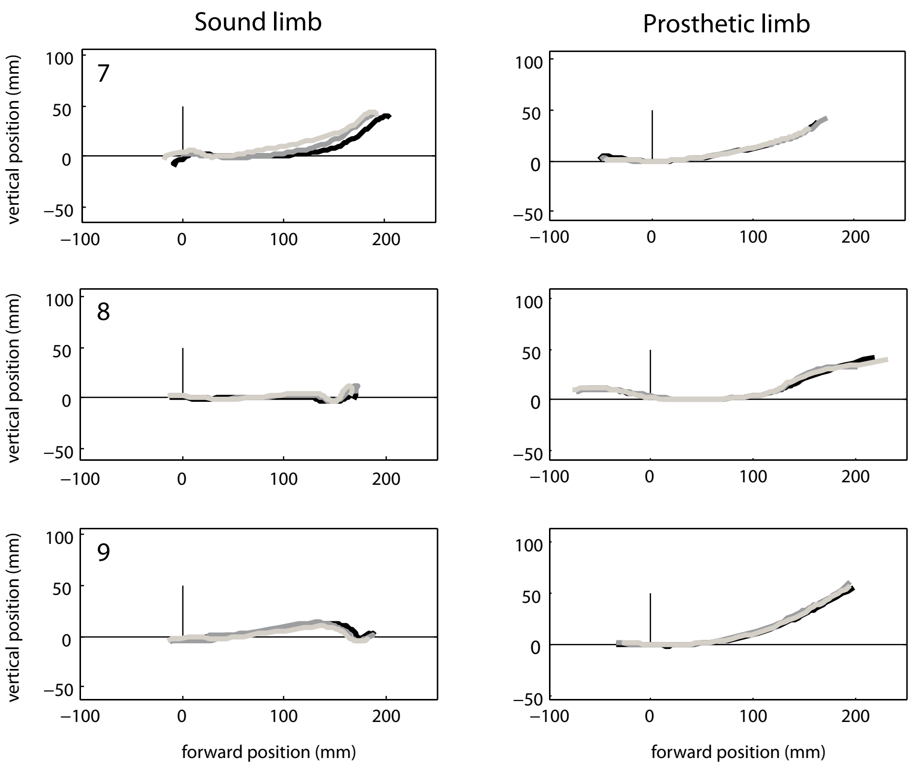

Figure 3.4 | Roll-over shapes of three trials for sound and prosthetic limbs in participants with transfemoral amputation (participants 7–9).

end (participants 2 and 5) of some of the roll-over shapes relate to a translation of the ankle with respect to the floor at the beginning or end of stance. Here, the stance limb supported more than 1/3 of body weight.

Discussion

In this single-case design study, we showed that ankle-foot roll-over shapes in people with amputation are highly individual. Most prominent were the adjustments of roll-over shape of the sound limb in the participants with transfemoral amputation. Each participant had his own unique individual roll-over shape, like a personal signature. In order to ensure ground clearance

Table 3.3 | Roll-over characteristics.

	Radius of curvature (mm)			Roll-over shape comparison (r.m.s. mm)		
	Sound/ Left	Prosthetic/ Right	Ratio	Reproducibility		Asymmetry
				Sound/ Left	Prosthetic/ Right	Left vs Right / Sound vs Prosthetic
Controls						
1	276 (8.3) [0.978 – 0.984]	1145 (242.4) [0.866 – 0.940]	0.25	1.59 (0.47)	6.20 (6.20)	18.07 (4.28)
2	288 (11.5) [0.984 – 0.994]	445 (14.9) [0.972 – 0.988]	0.65	5.44 (1.98)	5.20 (2.92)	28.93 (4.40)
3	357 (14.1) [0.987 – 0.992]	522 (39.3) [0.978 – 0.995]	0.69	7.35 (3.24)	3.09 (1.19)	35.83 (5.05)
Transtibial amputees						
4	263 (5.2) [0.941 – 0.972]	271 (17.2) [0.981 – 0.986]	0.97	3.84 (1.23)	10.97 (4.61)	14.13 (6.68)
5	528 (47.5) [0.993 – 0.996]	344 (7.8) [0.993 – 0.995]	1.53	10.57 (4.95)	5.47 (3.65)	38.64 (7.54)
6	466 (5.2) [0.940 – 0.996]	420 (23.4) [0.950 – 0.975]	1.11	4.31 (1.13)	4.05(2.61)	33.93 (3.01)
Transfemoral amputees						
7	309 (52.2) [0.930 – 0.972]	401 (33.3) [0.989 – 0.995]	0.77	9.83 (3.22)	4.02 (2.55)	41.05 (5.97)
8	95291* (159569.8) [0.171 – 0.245]	387 (32.1) [0.884 – 0.908]	255.13	2.40 (0.54)	8.20 (5.71)	10.37 (1.62)
9	1481* (2799.0) [0.204 – 0.457]	306 (11.5) [0.991 – 0.995]	4.84	19.10 (7.14)	18.28 (12.32)	24.56 (3.43)

mean (S.D.) [range of coefficient of determination]

* inaccurate due to strong deviation of the roll-over shape from circular

of the swinging prosthetic limb, some participants with transfemoral amputation lifted/pushed themselves up in late stance (van der Linden et al., 1999). Because of the passive properties of the prosthetic system, all compensations need to be made by the sound limb (Schmid et al., 2005). In steady-state walking, the roll-over shape of the prosthetic limb will be about the same in every step. The roll-over shape in the sound limb can be actively adjusted to the limitations of a particular prosthetic limb. These adjustments were different but very repeatable in each participant, indicating that they had all established a steady gait pattern. In the participants with transtibial amputation, we also found some interlimb differences; these were, however, considerably smaller.

Interestingly, even though the roll-over shapes generated during the stance phase of the prosthetic limb are comparable with those found in able-bodied controls, the participants with transfemoral amputation still adjusted their roll-over shape on the side of the sound limb. The three participants with transtibial amputation, however, appeared to benefit from the "natural" roll-over shape of the prosthetic limb. The radius of curvature in the prosthetic limb fell within the range of control values. The radii of the prosthetic limbs appeared to range between 30 and 43% of the leg length. Depending on the relative radius of curvature, a prosthetic foot can be considered as more or less stable (*Chapter 2*; Curtze et al., 2009). Previous research shows that able-bodied people walk with curvatures of about 30% of the leg length (Hansen et al., 2004a; McGeer, 1990). In combination with the remaining knee function in the amputated limb, this natural roll-over shape of the prosthetic limb allows people to achieve a symmetric walking pattern. The participants with transtiabial amputation showed little signs of adjustment in the sound limb, which had a more or less natural circular ankle-foot roll-over shape.

With respect to the interlimb differences, the three controls did show a less symmetric roll-over shape than expected. These high interlimb differences in the controls may be caused by functional gait asymmetry, where the non-dominant limb contributes more to support. The dominant limb is thought to contribute more to the push-off during propulsion (Sadeghi et al., 1997), which might be reflected in a more flattened roll-over shape. All three controls indicated their right limb as dominant, which agrees with the larger radii found for this side. This potential relation deserves further investigation in a larger population.

Finally, it should be noted that determining the radius of curvature by fitting a circular arc to roll-over shapes has its limitations. When applying this method, it is assumed that roll-over shapes are circular. However, as the cases presented here illustrate nicely, this assumption is not satisfied in all cases. The method is very sensitive to any deviations from circular, affecting the estimation of the radius of curvature strongly. Furthermore, this is reflected in a decrease of the coefficient of determination (R^2), a measure for how well the data are predicted by the model. In this study, we determined the radius of curvature by averaging over shape, hereby giving equidistant points equal weight. When nonresampled time data would have been used, regions where the CoP changes little with time (at initial and final stance) would have received even more weight. For not strictly circular roll-over patterns, this leads to strong deviations in the estimation of the radius. Nevertheless, radii found for shapes with a close-to-circular shape were approximately 30 to 50 % of the leg length, a range that was found to be metabolically optimal (Adamczyk et al., 2006).

In cases where the assumption of circularity of the roll-over shape is violated, statistical testing for differences in radius of curvature between limbs/subjects appears inappropriate. Rather, analyses should aim toward discussing the individual adjustments of such cases.

When determining asymmetry of roll-over shapes by means of radius of curvature, we should note that this measure is sensitive to deviations from circular shape, while when using the r.m.s. distance, the outcome is sensitive to shift between curves. Consequently, when comparing two identical curves, the r.m.s. distance can deviate from zero in case these curves are shifted. However, two curves can be different in shape although they have the same best-fit radius of curvature. As both measures highlight different aspects of asymmetry, they should be used complementarily instead of interchangeably. The analyses of temporal gait dynamics revealed a typical prolonged stance phase on the sound limb in participants with transfemoral amputation. This asymmetry in stance phase agrees with the findings of previous studies (Hof et al., 2007; Schmid et al., 2005). In the participants with transtibial amputation, this asymmetry was also present but less distinct (Vanicek et al., 2009a). Additionally, we found this asymmetry to be strongly influenced by the level of amputation. The three participants with transtibial amputation had a much more symmetric stance phase ratio compared with the three participants with transfemoral amputation.

Another characteristic finding is the prolongation of the double-support phase for the prosthetic limb in participants with transfemoral amputation (Schmid et al., 2005), which can be considered to increase stability during stance on the prosthetic limb. Again, the participants with transtibial amputation did not show this adjustment but had a quite symmetric double-support phase. It appears that both adjustments, the prolonged stance on the sound limb and the prolonged double-support phase on the prosthetic limb, correlate with the level of amputation. The loss of the natural knee joint and resulting absence of direct muscular joint control in participants with transfemoral amputation appears to be critical for an increase in stabilizing adjustments. However, we present the individual temporal gait characteristics purely to indicate that these cases are within the normal range of people with amputation.

Limitations of this study are the small study group size and limited number of walking trials per participant. However, despite the small sample size, the results show that roll-over shapes deviate from a circular shape as an effect of adjustments in the sound limb, thereby violating the generally assumed circularity (Hansen et al., 2003; Hansen and Childress, 2004; Hansen et al., 2004a; Hansen and Childress, 2005; Miff et al., 2008). Future research on roll-over shapes in prosthetics should give more attention to this aspect.

Conclusions

Patients with amputation often desire a symmetric gait pattern, and determining roll-over shape can help identify asymmetries. However, the amputation locomotor system is inherently asymmetric, and strong arguments exist against reestablishing a symmetric gait pattern without considering the functional consequences. Determining roll-over shapes might eventually help us better understand adjustment strategies in prosthetic walking and to achieve an optimal compromise for each individual patient.

Suppliers

- Advanced Mechanical Technology, Inc, 176 Waltham Street, Watertown, MA 02472-4800, USA
- Vicon Motion System, 14 Minus Business Park, West Way, Oxford, OX-2OJB, UK
- Aurion S.r.l., Viale Certosa 191, 20151 Milano, Italy
- The MathWorks, Inc., Crystal Glen Office Centre, 39555 Orchard Hill Place, Suite 280, Novi, MI 48375, USA

Conflict of Interest

The authors declare to have no conflict of interest in this work.

Acknowledgements

We thank Henk Zijlstra for inspiring this research project and Ronald Davidsz for his assistance with the data collection. Furthermore, the authors wish to thank Rainer Breitling for his helpful comments on the manuscript.

The Narrow Ridge Balance Test: A Measure for One-leg Lateral Balance Control

4

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Gait & Posture (2010) 32: 627–631

Abstract

The assessment of balance capacity for people with widely different balance abilities is an important issue in clinical practice. We propose the narrow ridge balance test as a sensitive tool to assess one-legged balance capacity. In this test, participants are asked to perform single-leg stance on ridges of gradually decreasing width (100, 80, 60, 40, 20, 10, and 4 mm). An outcome measure was developed, base on time in balance in relation to the gradually decreasing ridge width. To evaluate the sensitivity and discriminating power of the test, we applied it to two groups of participants, a group of young participants (age 20–30 years) and group of healthy elderly participants (age 60–80 years). The test showed to sensitively differentiate between the two groups, showing lower scores for the elderly. Furthermore, the test appeared to identify large within-group differences. A special feature of this setup is that the difficulty of the test increases with the balance capacity of the participant. In this way, each participant is exposed to the maximally challenging task, and a broader variety of balance control mechanisms come into play. Finally, the outcome score of the new test was contrasted to conventional measures of standing balance, showing good agreement.

Introduction

The determination of balance capacity is an important issue in clinical practice. Clinical balance tests (Berg et al., 1992; Mathias et al., 1986; Podsiadlo and Richardson, 1991; Tinetti, 1986) aim at identifying people with low postural stability and an increased risk of falling. However, we argue that at the same time it is equally important to identify those people who have excellent balance control and consequently make particularly good candidates for specific treatments or interventions. A practical example concerns prosthesis prescriptions in lower limb amputees where one-legged balance was found to be a relevant predictor of functional outcome (Schoppen et al., 2003). Therefore, it is crucial that a balance test is sensitive over a wide spectrum of balance performances, differentiating high-performers from low-performers. This is rarely the case for existing clinical tools for assessing balance capacity which are not discriminative enough to draw reliable conclusions about individual patients. The predictability of falls in community-dwelling elderly based on force plate measures of quiet standing was found to be poor (Brauer et al., 2000; Scott et al., 2007). A limitation of the prevailing balance tests is that they only target the most basic balance mechanism, i.e. the ankle strategy. However, preventing a fall requires a wider spectrum of balance mechanisms. Here, arm movements do not only have a protective function, as reaching or grasping for external support, but through counter-rotation movements of the arms they can help to prevent an impeding fall (Pijnappels et al., 2009). A better falls risk prediction was reported for more dynamic tests, such as rapid voluntary stepping (Brauer et al., 2000). Furthermore, the control of lateral balance appears to be closely related to fall risk (Hilliard et al., 2008; Lord et al., 1999; Maki et al., 1994; Stel et al., 2003). It appears from these studies that balance control is best determined in dynamic situations challenging lateral balance. In this study, we present a newly designed balance task that allows a flexible assessment of lateral balance capacity at increasingly demanding levels. To cover the wide spectrum of balance capacity, we here contrast the balance control of a group of young people to that of a group of elderly. Finally, we compare the narrow ridge balance test with the commonly used method for assessing balance capacity: quiet standing on a force plate.

Methods

Participants

Two groups of participants were included in this study (Table 4.1): one group of young participants (n=21; 20–30 years; university students), and one of elderly participants (n=17; 60–80 years; members of local bingo, bridge, and billiard clubs). All participants had normal or corrected-to-normal vision and were not known to suffer from disorders influencing balance capacity.

Table 4.1 | Participant data.

	Young (n = 21)	**Elderly (n = 17)**
Age (yr)	22.7 (1.8)	69.5 (6.2)
Body weight (kg)	71.2 (8.4)	81.6 (11.4)
Height (m)	1.77 (0.09)	1.71 (0.07)
Sex (male/female)	11/10	10/7

mean (S.D.)

Apparatus

Ridges of 25 mm height and different width were custom-made from timber and iron. The widths of the ridges are 100, 80, 60, 40, 20, 10, and 4 mm (Figure 4.1). The time in balance was recorded by means of a stopwatch.

The ridges were placed on a Bertec force plate type 4060, recording at a sample frequency of 100 Hz. The force plate signal is converted by a 16-bit A/D converter and processed on a PC with Matlab. For the computation of the center of mass (CoM) position from the center of pressure (CoP) data the "combi" method was used (Hof, 2005).

Procedure

Participants started with quiet standing on two feet for 30 seconds, they were instructed to stand with their feet together and arms hanging loosely along their side. Then, participants were asked to balance on one leg on ridges of

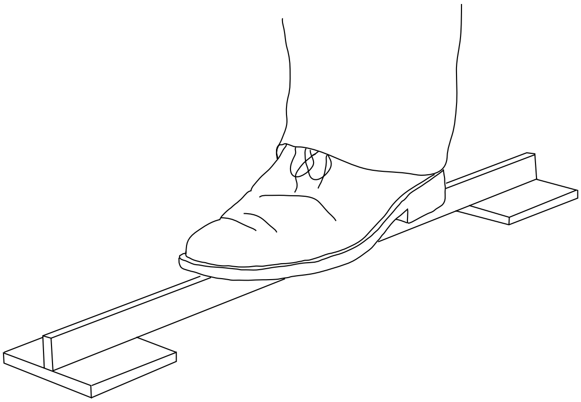


Figure 4.1 | Schematic illustration of the test setup (10-mm ridge).

gradually decreasing width. At the most basic level, participants performed single-leg stance on the floor. When they successfully keep balance for 20 seconds they proceeded to the next level, which was standing on a ridge of 100 mm width. Participants were instructed to maintain one-legged balance on the ridge, which was oriented in dorsoventral direction (Figure 4.1). The procedure of the test was guided by an "up-and-down" principle, i.e. if an individual was successful in keeping balance for 20 seconds, the ridge width was reduced; else the ridge width was increased. Trials on the narrowest ridge of 4 mm were not aborted after 20 seconds but the maximum time in balance was recorded. After five trials with less than 20 seconds in balance or more than 20 seconds on the narrowest ridge, respectively, the test was stopped. The benefit of this up-and-down principle is that the largest number of repeated measures will be taken at an individual's most challenging level. Participants were allowed to take rests between trials. Throughout the test, legs were tested alternatingly. Participants wore their habitual shoes with a rigid sole; wearing of sandals was not permitted. Before the test, foot-dominance was assessed by a four-item questionnaire (Coren, 1993).

Outcome Parameters

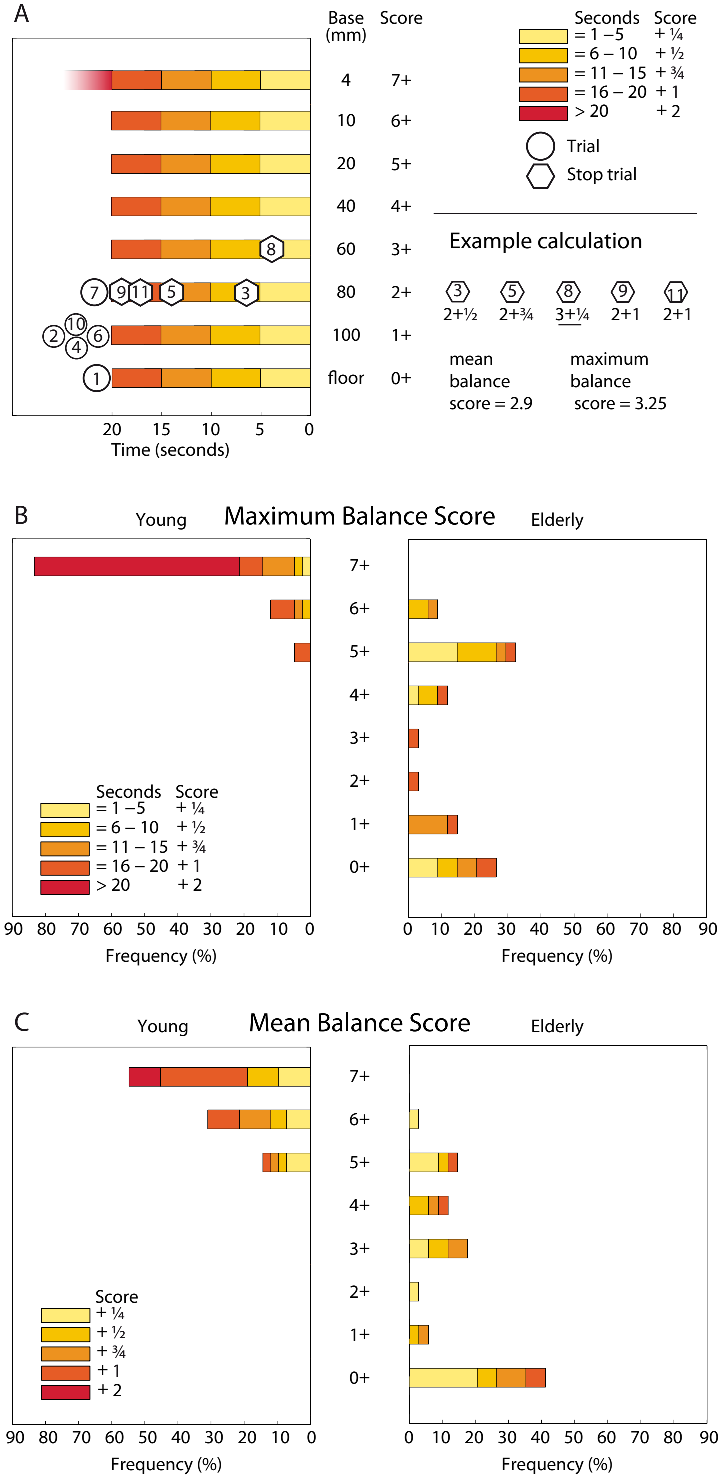
Time in Balance – Base of Support – Score

The principal outcome measure of the narrow ridge balance test is time in balance in relation to the width of the base of support. In order to compare and classify participants it is essential to condense this time and width information into a single parameter/quantity. For this purpose we developed a simple scoring scheme: every 1–5 seconds in balance earns ¼ point; on each level (different ridges including the floor) participants could earn up to one point (for 16–20 seconds in balance). Figure 4.2A illustrates the calculation scheme. In addition to this maximum score, based on the best trial per leg, we calculated a more robust estimate of balance performance, the mean balance score, by averaging the scores of the five "stop trials", i.e. those five trials in which participants stepped off the ridge ahead of time. Both the mean and maximum balance scores are calculated for each leg individually.

Mechanisms for Balance Control

For standing balance a number of motor strategies have been described, which can be classified into three mechanisms (Hof, 2007). In the first mechanism the CoP is moved by muscle action, classically referred to as "ankle strategy",

Figure 4.2 | (A) Calculation scheme for the balance scores. In the given example, the participant was successful in maintaining balance for 20 seconds on the floor and on the 100 mm ridge (trial 1 and 2). In trial 3 the participant stepped off the 80 mm ridge after 6 seconds, after which he again balanced successfully on the 100 mm for 20 seconds (trial 4). This 'up-and-down' procedure between the different ridge widths continued until the total of five stop trials (trial 3, 5, 8, 9 and 11) was reached. The maximum balance score was achieved during trial 8 on the 60 mm ridge, on which the participant maintained balance for 4 seconds; this leads to a maximum balance score of 3.25. The mean balance score was calculated as the mean of the scores of all five stop trials, yielding an average of 2.9. **(B)** Maximum balance score. Distribution of maximum balance scores (best trials) for the groups of young and elderly participants. Each row corresponds to a group of participants which had their best trial on a ridge of a particular width. The rows are further subdivided according to how long the participants managed to balance on this most challenging ridge. **(C)** Mean balance score. Distribution of mean balance scores for the groups of young and elderly participants. The subdivision is equivalent to panel (B).


A
Base (mm)
Score
4 7+
10 6+
20 5+
40 4+
60 3+
80 2+
100 1+
floor 0+
20 15 10 5 0
Time (seconds)
Seconds Score
= 1 – 5 + ¼
= 6 – 10 + ½
= 11 – 15 + ¾
= 16 – 20 + 1
> 20 + 2
Trial
Stop trial
Example calculation
2+½ 2+¾ 3+¼ 2+1 2+1
mean balance score = 2.9
maximum balance score = 3.25
B
Young
Maximum Balance Score
Elderly
Frequency (%)
C
Young
Mean Balance Score
Elderly
Score
Frequency (%)

in which the muscles around the ankle act to control balance. Next to this the "counter-rotation" mechanism was identified, which covers among others the "hip strategy". The third mechanism refers to external support, which was not allowed in this experiment. When standing with one leg on a broad support, the main strategy is the lateral ankle strategy (Hoogvliet et al., 1997; King and Zatsiorsky, 2002). Its mechanism is (lateral) displacement of the center of pressure (y_{CoP}): the acceleration of the CoM is proportional to the distance between CoP and CoM. When only the ankle strategy is used, the r.m.s. (root-mean square) value of this difference (y_{CoP} - y_{CoM}) is an indication for the extent to which the lateral ankle strategy is used. In addition, we determined the velocity of the lateral CoP displacement (v_{yCoP}).

When standing on a narrow support, a second mechanism comes into play, the "counter rotation" mechanism. Examples of strategies related to this mechanism are the hip strategy (Horak and Nashner, 1986), arm rotation and rotation of the free leg. Its magnitude can be represented by the r.m.s. value of a quantity y_h, which can be obtained by measuring the horizontal CoM acceleration with the force plate and subtracting the contribution of the first mechanism (Hof, 2007):

$$y_h = (y_{CoP} - y_{CoM}) + l\,\frac{F_{Gy}}{F_{Gz}}$$

in which l is the effective pendulum length (trochanteric height times 1.34; Massen and Kodde, 1979), multiplied by the proportion of medio-lateral and vertical component of the ground reaction force $\boldsymbol{F_G}$.

Statistical Analysis

As the balance score data were not normally distributed (Kolmogorov–Smirnov test) nonparametric tests were applied. Group differences were determined by the Mann-Whitney *U* test. Wilcoxon signed-rank tests were performed to test the agreement between balance scores of the left and right leg. In addition, we controlled for laterality (Mann-Whitney *U* test), grouping the participants with respect to their leg dominance, the dependent variable was the difference in balance score between left and right. The significance level was set to $p \leq .05$.

To evaluate the relationship between the two outcome scores, mean and maximum balance score, a linear regression analysis was performed. Furthermore, linear regression models were built to determine the association between the mean balance score and the different balance control parameters.

Results

Balance Scores

The narrow ridge test was evaluated with respect to maximum and mean performance. The maximum balance score distribution of elderly participants (Figure 4.2B) shows two peaks at floor level (maximum balance score 0+) and on the 20-mm ridge (maximum balance score 5+). 26% of the elderly were not able to maintain balance for more than 20 seconds on one leg on the floor (maximum balance score ≤ 1), 9% not even for 5 seconds (maximum balance score ¼). The narrowest ridge (4 mm) was reached by none of the elderly participants (maximum balance score < 7). The young participants performed at a considerably higher level; 62% had their performance maximum on the narrowest ridge (4 mm), on which they maintained balance for more than 20 seconds (maximum balance score = 9). All young participants had at least one attempt on the 20 mm ridge or narrower. That is, they kept balance for more than 20 seconds on all broader ridges.

For the mean balance score, the elderly showed scores between one and seven points, while the young reached scores between five and nine points (Figure 4.2C). 21% of the elderly have a mean balance score of ≤ ¼ points, i.e. on average, they maintain one-legged stance on the floor for less than 5 seconds. There was a strong relation between both scores; the mean balance score was explained to 94.9% by the maximum balance score ($R^2 = .949$, $p < .001$).

The narrow ridge test revealed significant group-differences (mean balance score: $U = 0$, $n_1 = 21$, $n_2 = 17$, $p < .001$; maximum balance score: $U = 0$, $n_1 = 21$, $n_2 = 17$, $p < .001$). There were no significant differences in performance between the left or right leg (mean balance score: $Z = -.868$, $p = .385$; maximum balance score: $Z = -1.120$, $p = .263$). Furthermore, leg dominance did not reach statistical significance (mean balance score: $U = 65.5$, $n_1 = 7$, $n_2 = 31$,

p = .098; maximum balance score: U = 82.5, n_1 = 7, n_2 = 31, p = .327).

The time to administer the test was strongly dependent on participant's balance capacity. For the elderly it took about 8 minutes to administer the test, while for the young it took about 15 minutes.

Balance Control Parameters vs. Mean Balance Score

Linear regression analyses of the lateral CoP deviation in standing on two feet and the mean balance score (averaged over left and right limb) revealed only a low, but statistically significant relationship (R^2 = .215, p = .005). Slightly stronger associations were found for the r.m.s. y_{CoP} of one-legged standing, a conventional measure for stability (Prieto et al., 1996), and the mean balance score (R^2 = .373, p < .001; Figure 4.3A), as well as for r.m.s. v_{yCoP} of one-legged standing and the mean balance score (R^2 = .436, p < .001; Figure 4.3B). Both r.m.s. y_{CoP} and v_{yCoP} increase with a lower balance scores; see Figure 4.3.

In Figure 4.4 r.m.s. y_{CoP} and y_h are given as a function of the base of support. At floor level r.m.s. y_h is lowest, in the best group (mean balance score > 7) even below r.m.s. y_{CoP} (Figure 4.4). This means that in the best group there is virtually no arm, leg or trunk motion in one-legged standing. At ridge widths smaller than 60 mm r.m.s. y_h starts to increase, to reach a level of 40–50 mm on

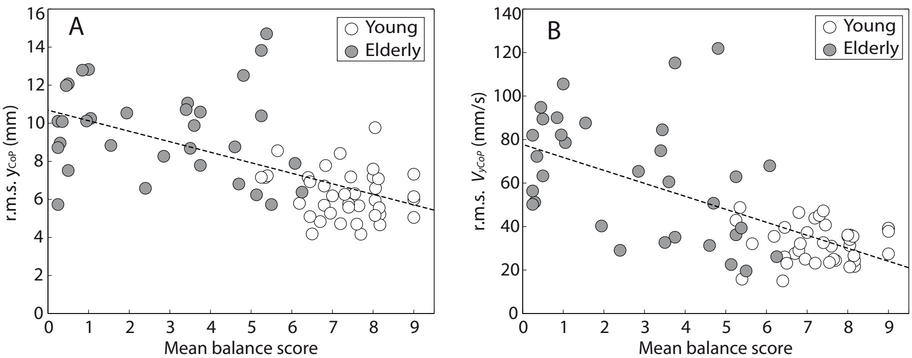


Figure 4.3 | (A) Root-mean square values of y_{CoP}, the lateral CoP deviation, and **(B)** v_{yCoP}, the velocity of lateral CoP deviation in one-legged standing versus the mean balance score for the group of young and elderly. The linear regression is indicated by the stippled line.

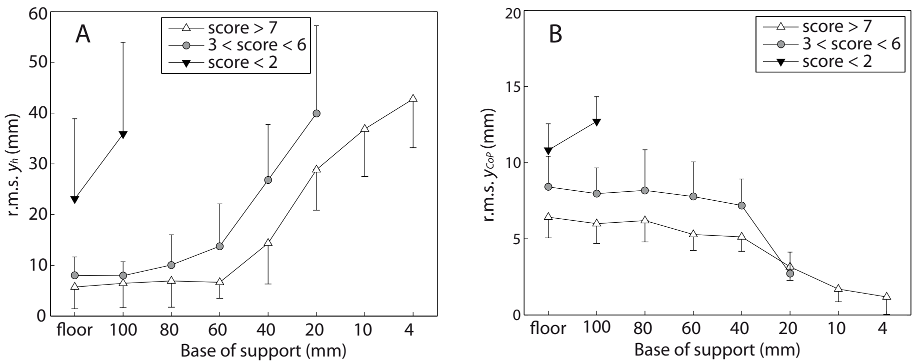


Figure 4.4 | (A) Root-mean square values of y_h, proportional to arm and leg motion, and **(B)** y_{CoP}, motion of the CoP under the foot, as a function of the base of support. Data of three groups are presented: The best group (*n* = 13) had scores above 7, the midscore group (*n* = 11) had scores between 3 and 6, and the worst group (*n* = 8) had scores below 2.

the narrowest ridge. The lateral CoP movement r.m.s. y_{CoP} decreases with ridge width, as it is constrained by the base of support. At the narrowest ridge an individual participant can stand on, the value of y_h is around 40–50 mm r.m.s. for all participants, irrespective of their balance score.

Discussion

In the present study, we introduce a new balance task. By gradually reducing the width of the base of support, different balance control strategies come into play (Hof, 2007; Otten, 1999), making this balance test unique in its ability to discriminate performance across a wide range of abilities: both young and elderly show a wide spread of individual scores. Due to the gradually increasing difficulty of the task, high performers are highly challenged, while the high-risk persons are not overstrained by balancing on a too narrow ridge. The narrow ridge balance test appeared to be a strict measure for lateral balance control, identifying even small differences between individuals, which might prove advantageous with respect to the sensitivity of the test in future studies. The observed ceiling effects for very high-performers can be considered as a minor problem, as a more fine-graded delimitation in this group of young, healthy people, who are endowed with excellent balance control, was not an interest of this study. Furthermore, this ceiling effect is significantly reduced through calculating the mean balance score instead of the maximum balance score. Naturally there is a corresponding floor effect for the very frail who are unable to keep balance on one leg even on level ground and thus are best assessed by alternative balance tests (Berg et al., 1992; Mathias et al., 1986; Podsiadlo and Richardson, 1991; Tinetti, 1986).

In this study good agreement was found between the mean balance score and both the r.m.s. y_{CoP} and the r.m.s $\boldsymbol{v}_{yCoP}$ in one-legged standing, which are common measures for stability (Prieto et al., 1996). With decreasing ridge widths, participants gradually switched from controlling balance by "moving the CoP" to "counter-rotation". Participants with low balance scores showed higher r.m.s. y_h at the wider ridges already, which was visible in vigorous arm and leg motions. Furthermore, our data suggests a boundary for counter-rotation (r.m.s. y_h) around 40–50 mm, which agrees closely with Otten (1999), who reported maximum values of up to 60 mm for balancing on a 4-mm ridge.

With the current test protocol it took about 8–15 minutes to administer the test. By modifying the protocol this administration time can be reduced significantly: first, by reducing the amount of ridges, and second, from our results it appears sufficient to test subjects on one leg only.

Our balance test is proposed as a useful instrument for clinical practice. The excellent correlation between the robust mean balance score and the easy-to-calculate maximum balance score indicates that the up-and-down principle leads to a reliable balance assessment. An added advantage for practical applications is its simplicity: no intricate apparatus is needed to measure sway. The narrow ridge balance test is designed to fill the current gap for tools that can be used to assess balance over a wide spectrum of balance capacities, thereby addressing different balance control mechanisms. Ongoing research aims at studying the narrow ridge balance test's test-retest reliability and its validity for different (patient) populations. Future work should examine the degree of external as well as internal validity of this new measure.

Suppliers

- Bertec Corporation, 6171 Huntley Road, Suite J, Columbus, OH 43229, USA
- The MathWorks, Inc., Crystal Glen Office Centre, 39555 Orchard Hill Place, Suite 280, Novi, MI 48375, USA

Acknowledgments

The authors wish to thank Rainer Breitling for his helpful comments on the manuscript.

The Relative Contributions of the Prosthetic and Sound Limb to Balance Control in Unilateral Transtibial Amputees

5

Carolin Curtze, At L. Hof, Klaas Postema & Bert Otten

submitted

Abstract

In unilateral transtibial amputees maintenance of standing balance is compromised due to the lack of active ankle control in the prosthetic limb. The purpose of this study is to disentangle the contribution of the prosthetic and sound limb to balance control following waist-pull perturbations. We compared the contribution of the hip and ankle joints to balance control of 15 unilateral transtibial amputees and 13 able-bodied controls after been externally perturbed through release of a pulling force. Perturbations were applied in four different directions. Outcome measure was the proportion of joint moment integrated over time generated by the hip and ankle joints in order to restore static stability after perturbation. Analyses revealed that perturbations in backward/forward direction were recovered mainly by the ankle strategy. The amputees compensated for the absence of active ankle control in the prosthetic limb by increasing the ankle moment in the sound limb. Interestingly, the passive properties of the prosthetic foot contributed to balance control. Amputees and controls resisted perturbations in medio-lateral direction by generating the necessary hip moments. Finally, these findings are discussed with respect to prosthetic design and rehabilitation processes.

Introduction

While standing in highly frequented places, we oftentimes are subject to perturbations caused by others inadvertently coming into contact with us. Here, balance is threatened as perturbations applied to the upper body will accelerate the body's center of mass (CoM) towards the boundary of the base of support. In order to withstand these perturbations without losing balance, we can use different balance control strategies. Depending on the magnitude of the external perturbation, balance can be restored by means of the ankle and/or hip strategy, or the load-unload strategy (Winter et al., 1996); large perturbations will require a protective stepping strategy (*Chapter 6*; Curtze et al., 2010a). When standing in perfect equilibrium the CoM is vertically above the center of pressure (CoP), and the ground reaction force ($\boldsymbol{F_G}$) is pointing towards the CoM (Figure 5.1). After being perturbed by a horizontal force, such as a push or pull, it is important to generate an equivalent force in opposite direction; this can either be done by the hip or the ankle strategy or a combination of both strategies. The direction of $\boldsymbol{F_G}$ can be modulated by generating a hip moment (M_H), thereby moving the CoM. In so doing, a horizontal force (F_{HM}) is produced

$$F_{HM} = M_H / L$$

in which L is the leg length, i.e. the trochantic height.

In the ankle strategy, the position of the CoP under the feet is moved by means of muscles action around the ankle joint. When only the ankle moment (M_A) is acting the horizontal component (F_{AM}) of $\boldsymbol{F_G}$ is

$$F_{AM} = M_A / H$$

in which H is the height of the CoM.

To compare the amount of ankle and hip strategy used, it will be assumed that $H \approx L$, which is approximately true for most humans. With this approximation the horizontal component of $\boldsymbol{F_G}$ due to hip or ankle F_{HM} and F_{AM} are both proportional to the joint moments M_H and M_A. This allows a simple comparison of the amount of ankle and hip strategy used.

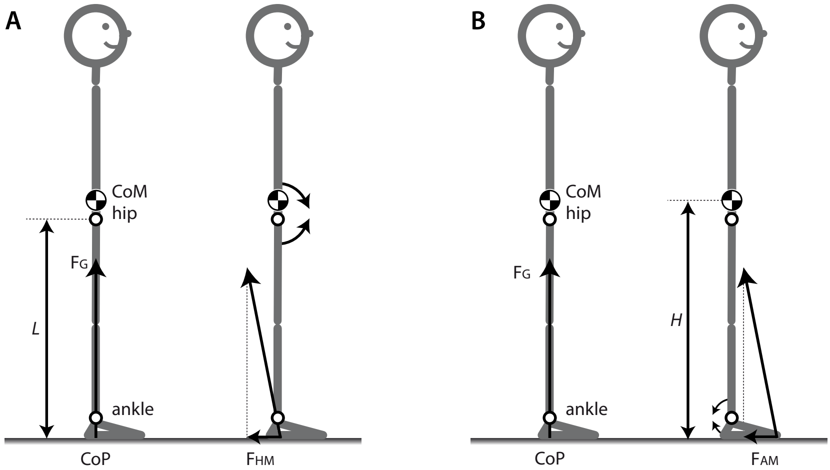


Figure 5.1 | Hip and ankle strategy. Quiet standing in perfect equilibrium versus **(A)** hip strategy, and **(B)** ankle strategy.

Note that Figure 5.1 and the principle outlined above are based on the choice of the position of the ankle and hip joints being aligned with the ground reaction force. This in reality is usually not the case (for instance the CoP is often in front of the ankle joint), but the principle of horizontal force changes still holds when offset moments of force are required for static standing.

Pioneering work by Horak and Nashner (1986) has shown a dominance of the ankle strategy during small perturbations such as brief horizontal platform translations in anterior-posterior direction. In the medio-lateral direction the position of the CoP is controlled by the hip abductors/adductors working in antiphase, i.e. the load-unload strategy. During quiet standing the hip abductors/adductors moments are equal in magnitude, so that their effects cancel (Winter et al., 1996).

For lower limb amputees external perturbations are a particularly challenging, as their balance is impaired due to the absence of active muscular control of one or more joints. An approach to study the effect of external perturbations on quiet standing is to apply sudden pulls to the waist (Luchies et al., 1994; Pai et al., 1998; Mille et al., 2003; Hsiao-Wecksler et al., 2003). In previous research this task has been used in able-bodied subjects to investigate the boundaries of protective stepping as a response to sudden waist-pull perturbations (Luchies et al., 1994; Pai et al., 1998; Mille et al., 2003). In the present study only mild perturbations will be applied which can be withstood without stepping. Studying the response of unilateral transtibial amputees appears particularly interesting as these individuals lack the active control of the prosthetic ankle joint, while having two intact hip joints and one intact ankle joint. In this task, when standing with extended knees, the knee joints don't contribute to balance control. The aim of this study is to disentangle the contribution of the prosthetic and sound limb/joints to balance control. We hypothesised that unilateral transtibial amputees compensate for the lack of active ankle control in the prosthetic limb by increasing the contribution of the sound limb to balance control (Vrieling et al., 2008d). A study on the corrective ankle moments in lower limb amputees in response to platform perturbations revealed that the stiffness of the prosthetic ankle contributes to balance control (Nederhand et al., 2011). Perturbations in the sagittal plane were expected to result in an increase of ankle moment in the sound limb compared to the prosthetic limb or an able-bodied limb. No differences in the contribution of the prosthetic and sound limb were expected for perturbations in the frontal plane, as the load-unload strategy is acting and the hip joints of transtibial amputees are muscle empowered, just like in able-bodied controls. A better understanding of the dynamics of balance control will help to improve prosthetic design and rehabilitation processes.

Simulation

To validate the above mentioned assumption of $H \approx L$, a simulation of a simple planar mechanism of two elements constrained by an endpoint is performed (Otten, 2003). The elements are connected to each other with a hinge joint. The human body modeled is a person of 1.80 m tall and 80 kg body weight. The mass and length data of the head, arms, and trunk (HAT) segment and the leg segment were based on Winter (1990). It turns out that a hip moment of -100 Nm produces 162 N horizontal $\boldsymbol{F_G}$. Applying a moment of the same magnitude at the ankle produces 129 N horizontal $\boldsymbol{F_G}$. The separate contributions add up to a combined effect of 291 N horizontal $\boldsymbol{F_G}$. The hip moment contributes 55% and the ankle moment 45% of the total horizontal $\boldsymbol{F_G}$ at equal joint moments.

Methods

Participants

A group of 15 male unilateral transtibial amputees and 13 able-bodied controls were included in the study. The amputees had a mean age of 55.1 ± 9.8 years, height of 1.83 ± 0.52 m and weight 92.5 ± 13.9 kg. The most frequent reason of amputation was trauma (10), followed by vascular disease (4), and limb deficiency (1). Five of the 15 amputees had undergone a transtibial amputation of their right limb. The median time since amputation was 7 years (range 2–44 years). All amputees were experienced and able walkers. In the experiment the amputees used their habitual prostheses which were equipped with one of the following prosthetic feet: Otto Bock 1C40 (7), Otto Bock 1D10 (3), Otto Bock 1D35 (2), Otto Bock 1A30 (2), and Endolite Multiflex (1). The controls had a mean age of 53.1 ± 10.6 years, height of 1.87 ± 0.56 m and weight of 87.2 ± 10.1 kg.

Apparatus

The ground reaction forces were measured by two AMTI force plates, sampled at a rate of 1000 Hz (Figure 5.2). The kinematics were tracked by an eight-camera Vicon motion capture system sampled at a rate of 100 Hz. A total of 35 reflective markers were placed on anatomical landmarks as specified in the Vicon Plug-in Gait full-body model. Markers placed on the prosthetic limb matched the marker positions on the sound limb. Finally, anthropometric measurements were taken according to the requirements of the model, i.e. body mass, body height, leg length (anterior superior iliac spine to the medial malleolus), knee width (flexion axis), and ankle (medial and lateral malleoli) width for each of the limbs.

The setup for the perturbation consisted of a cable running over a pulley which was attached to a force transducer (Figure 5.2). The load was connected to the cable by means of an electromagnet. With a switch the current to the electromagnet could be interrupted, thereby releasing the load.

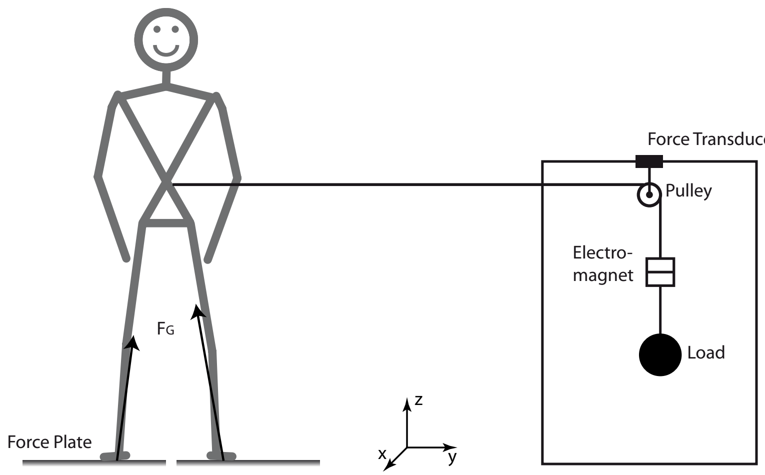


Figure 5.2 | Experimental setup. The participant resists a force pulling from the left. The electromagnet was used as release mechanism. Through release of the load a fall was induced opposite to direction the pulling load. The reading of the force transducer was used to determine the moment of release of the load. The participant was positioned on two force plates. The setup was equivalent for all four perturbation conditions.

Procedure

The participant stood upright with their feet parallel. Each foot was positioned on a separate force plate. A horizontal force was applied to his waist via a cable as described above (Figure 5.2). The participant was resisting the load before it was released by the experimenter at an unpredictable moment, inducing a fall opposite to the pulling direction. The participant had to compensate for that perturbation. Falls were induced in four different directions: (A) backward, (B) forward, (C) towards the sound/right limb, and (D) towards the prosthetic/left limb, by releasing a load pulling in opposite direction. The order of perturbations was randomized over participants; three trials were performed per direction. The load was relative to the participant's body weight: 2.5% in anterior-posterior, and 5% in medio-lateral direction. The participant's body weight (including the prosthesis) was determined beforehand. The lateral distance between the medial edges of the feet was set to 20% the height of application of the pulling force, which is about hip width. The participant was verbally instructed to resist the perturbation, without performing a step. The participants were secured by a safety harness system. The study was approved by the local Medical Ethical Committee.

Data analysis

The moment of release (t_0) was determined by means of readings from the force transducer. The return of static stability/balance (t_1) was defined as the moment at which the ground reaction force component in the direction of pull remains within ± 1 S.D. of the mean of static stability/balance, i.e. quiet standing without the load applied.

Finally, to disentangle the contribution of prosthetic and sound limb to balance control, the hip and ankle moment of each of both limbs were determined in the direction of perturbation. In the anterior-posterior perturbation condition, an increase in moment indicates plantar/hip flexion. In the mediolateral perturbation conditions an increase in moment in the limb ipsilateral to the fall indicates ankle inversion/hip abduction, while in the limb contralateral to the fall a decrease in moment indicates ankle inversion/hip abduction. Note that the limbs have opposite signs for ankle inversion/hip abduction, as their effects cancel.

Before determining the normalized joint moment integral generated by each of the four joints, the time-series data of the normalized moments (Nm/kg) were shifted by their baseline value, i.e. set to zero for t_1. This value corresponds to the equilibrium value for quiet standing without the load applied, which was used as reference point. By numerically integrating the normalized moment over time (t_0 – t_1), the normalized joint moment integral (Nm.s/kg), was determined. Subsequently, the contribution of each of the joints was expressed as a proportion of the total joint moment integral of all four joints.

Statistical analysis

For each of the perturbation conditions, the mean proportion of joint moment impulse was calculated for each of the four joints. The mean value was calculated from multiple trials from the same participant. Differences in outcome between the two groups were tested by one-way ANOVAs. Paired *t*-tests were used to compare the contribution of the ankle/hip joint of the left/prosthetic and right/sound limb to balance control. The significance level was set to $p \leq .05$.

Results

Exemplary time series data of the normalized ankle and hip moments for all four perturbation conditions are shown in Figure 5.3. The data are set to zero for t_1, which allows a better comparison of the contribution of each of the joints. In Figure 5.3A, the amputee resists the forward pull by the load, which is released at t_0. The sum of normalized moments (M_{sum}) reveals that it takes about 100–200 ms before the actual balance recovery response is initiated, i.e. drop in M_{sum} after release. In this short time window the participants starts to fall backwards. It is mainly the ankle moment generated by the sound limb (M_{AS}) and the passive properties of the prosthetic ankle (M_{AP}) that appear to contribute to balance recovery, while the hip moments of the sound (M_{HS}) and prosthetic (M_{HP}) limb seem to contribute less. In the late phase of balance recovery, about 1.5 s after the release of the load, there is a small overshoot of M_{sum}, before the amputee is in static balance at t_1. Note that M_{sum} is not the total moment of force acting on the body, but the sum of the moments of the separate joints is directly related with the balance recovering ground reaction force evoked.

By determining the proportion of joint moment integral the contribution of each joint to balance control can be dissociated. The proportion of joint moment integral for the amputees and controls for each of the perturbation conditions is given in Figure 5.4. In the amputees the sound (53%) ankle contributed significantly more to balance control after a fall induced in backward direction than the prosthetic (37%) ankle ($t(14) = 2.197$, $p = .045$; Figure 5.4A), while there was no difference in contribution between the hip of the prosthetic (6%) and sound (4%) limb ($t(14) = .223$, $p = .826$). In the controls the right (41%) and left (45%) ankle contributed equally to balance control after a fall was induced in backward direction ($t(12) = -.845$, $p = .414$). Furthermore, there was no significant difference in contribution of the right (5%) and left (9%) hip ($t(12\ ) = -.563$, $p = .584$). The proportion of joint moment integral of the ankle (43%) was significantly higher than that of the hip (7%; $t(12) = 11.143$, $p < .001$), which shows that the controls used an ankle strategy to recover balance. Between-group comparison revealed a significantly higher proportion of joint moment integral in the sound (53%) ankle of amputees than that in able-bodied controls (43%; $F(1,26) = 4.806$, $p = .038$).

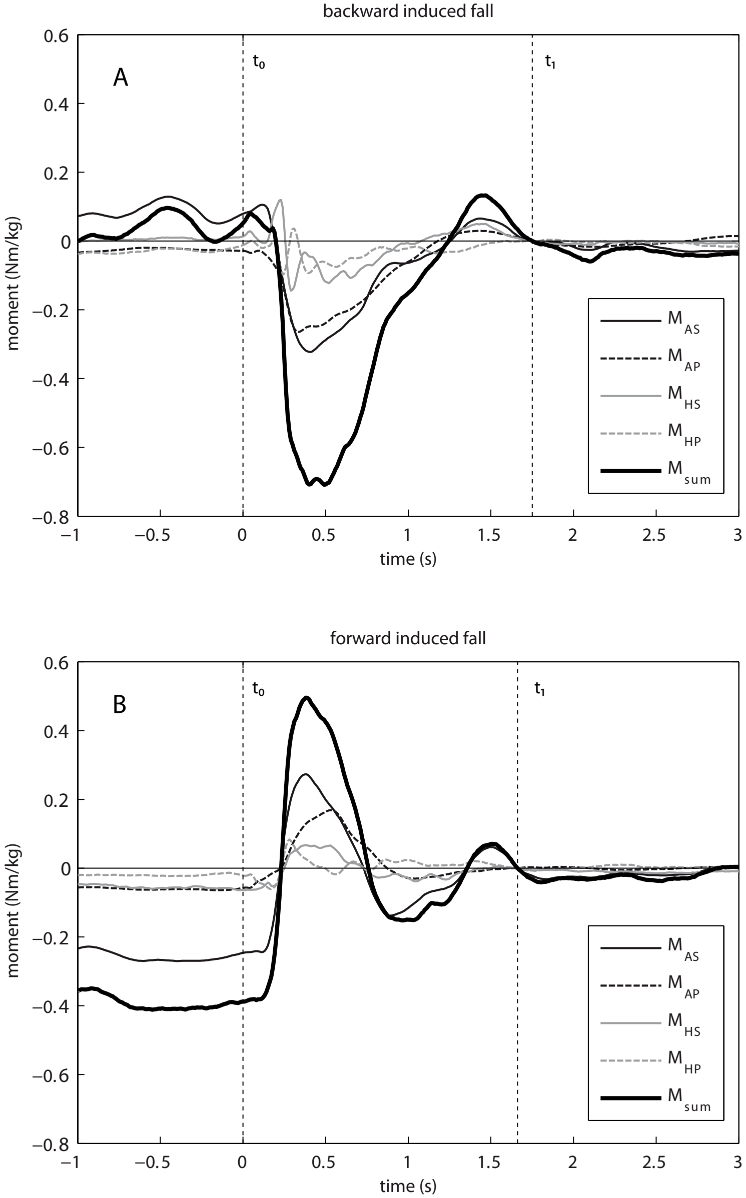


Figure 5.3 | Normalized ankle and hip moments in the direction of perturbation as a function of time. Exemplary normalized ankle moments of the sound (M_{AS}) and prosthetic (M_{AP}) limb, normalized hip moments of the sound (M_{HS}) and prosthetic (M_{HP}) limb, and the sum of the normalized moments (M_{sum}) are given for a single amputee participant. A fall was induced **(A)** backward, **(B)** forward, **(C)** towards the sound limb, and **(D)** towards the prosthetic limb by releasing a load that was pulling in opposite direction. *(continued)*

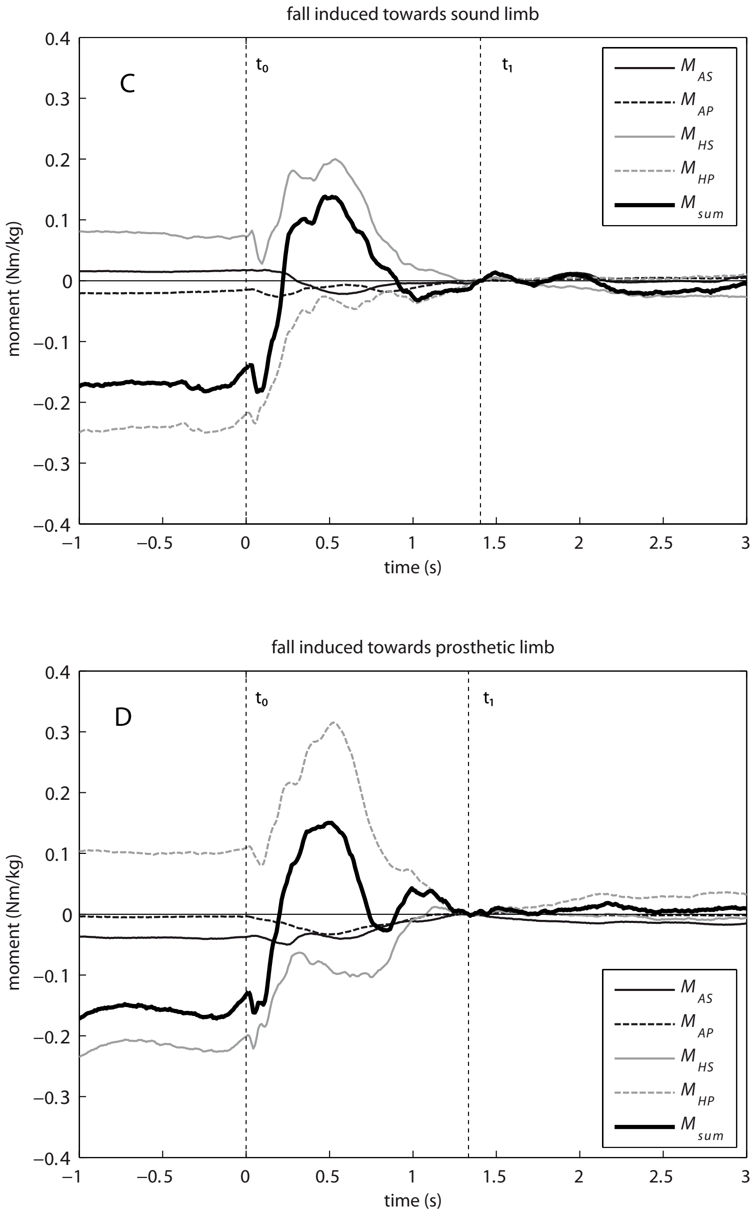


Figure 5.3 | *continued.* At t_0 the load was released, static balance was restored at t_1. All moments are set to zero for t_1. In (A, B) an increase in moment indicates plantar flexion/hip flexion. In (C, D) an increase in moment in the limb ipsilateral to the fall indicates ankle inversion/hip abduction, while in the limb contralateral to the fall a decrease in moment indicates ankle inversion/hip abduction. Note that the limbs have opposite signs for ankle inversion/hip abduction, as their effects cancel.

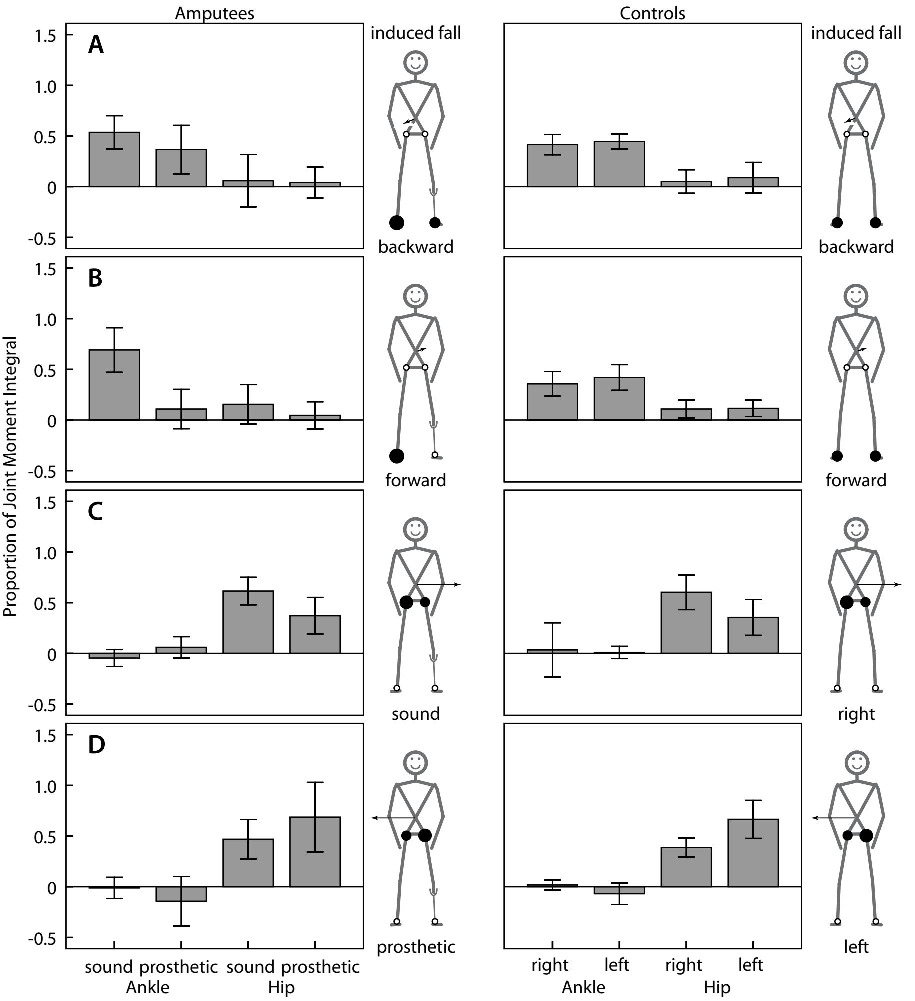


Figure 5.4 | Joint moment integrals as proportion of total for four different load release conditions: A fall was induced **(A)** backward, **(B)** forward, **(C)** towards the sound/right limb, **(D)** towards the prosthetic/left limb, by releasing a load that was pulling in opposite direction. Error bars depict ± 1 S.D.

The induced forward fall depicted in Figure 5.3B shows the comparable pattern as for the backward fall, but in direction reversed. In the initial phase, before release of the load, the amputee is resisting the backward pull of the load by means of the ankle moment generated by the sound limb. In this case, the prosthetic (11%) ankle appears to contribute less to balance recovery ($t(14) = 6.504$, $p < .001$; Figure 5.4B); the amputees were greatly relying on their sound ankle (69%). The controls, however, used their right (36%) and left (42%) ankle equally ($t(12) = -.985$, $p = .344$), and their right (11%) and left (12%) hip joints equally as well ($t(12) = -.167$, $p = .870$). Further analysis revealed that the ankle strategy was predominant in the control group ($t(12) = 10.813$, $p < .001$). Marked between-group differences were found for the proportion of joint moment integral of the ankle. While the sound (69%) ankle of amputees contributed significantly more to balance control than the ankle (39%) in able-bodied controls ($F(1,26) = 23.672$, $p < .001$), the prosthetic (11%) ankle contributed significantly less ($F(1,26) = 25.627$, $p < .001$).

In the medio-lateral perturbation conditions, Figure 5.3C-D, the amputee achieved stable balance by means of the load-unload strategy. Thus, balance is accomplished by the hip abductor/adductor moments and virtually no contribution of the ankle joints. In the early response, after release, the hip abduction moment in the limb proximal to the pulling load is reduced, while the hip abduction moment in the limb contralateral to the pulling load is increased before it finally converges towards its baseline value.

When a sideward fall was induced to the sound limb, by releasing a load to the side of the prosthetic limb, the sound (61%) hip joint contributed significantly more to balance recovery than the hip joint of the prosthetic (37%) limb ($t(14) = 3.145$, $p = .007$; Figure 5.4C). Furthermore, there was a significant difference in joint moment integral generated by the prosthetic (6%) and sound (-4%) ankle joint ($t(14) = -2.640$, $p = .019$). When the fall was induced to the prosthetic limb, by releasing a load to the opposite side, no significant difference was found with respect to the contribution of the hip joints of the sound (47%) and prosthetic (68%) limb ($t(14) = -1.734$, $p = .105$; Figure 5.4D). Moreover, there was no significant difference in contribution between the prosthetic (-14%) and sound (-1%) ankle joint ($t(14) = 1.949$, $p = .072$).

In the controls, the hip joint contralateral (63%) to the pulling load contributed significantly more to balance recovery than the proximal (37%) hip joint ($t(12) = 4.233$, $p = .001$; Figure 5.4C-D). No differences in proportion of joint moment integral were found for the ankle joints ($t(12) = -.586$, $p = .569$).

Discussion

In this study we disentangled the contribution of the prosthetic and sound limb to balance control in unilateral transtibial amputees following waist-pull perturbations. We found that perturbations in the anterior-posterior direction are recovered by the ankle strategy, while the hips play only a subordinate function, i.e. they contributed to a much lesser extent. In the induced backward fall condition the passive properties of the prosthetic foot contributed significantly to balance recovery in amputees; yet there was considerable compensation by the ankle of the sound limb. Horak and Nashner (1986) demonstrated that healthy people, when being perturbed while standing on a support surface shorter than the foot length, compensate for the loss of ankle strategy by using the hip strategy. Yet, the amputees in our study preferred to increase the contribution of the sound limb. In the induced forward fall condition the sound ankle compensated almost fully for the lack of active ankle moment in the prosthetic limb. This compensation effect agrees with the control strategies used by unilateral lower limb amputees when balancing on a moving platform (Vrieling et al., 2008d). The results of this study showed that most adjustments strategies in amputees occurred in the sound limb, under which they increased CoP movement and loading (Vrieling et al., 2008d). In another platform perturbation study it was further shown that the contribution of both ankles to balance control was even more asymmetric than the asymmetry in loading (Nederhand et al., 2011). Besides, the study confirmed the contribution of the passive ankle properties to balance control in anterior-posterior direction (Nederhand et al., 2011). The finding has important implications for prosthetic design and fitting. Zmitrewicz et al. (2007) have shown that a passive prosthetic foot-ankle device is functionally similar to the soleus muscle. Amputees with good muscle strength and control in the sound ankle can compensate for the lack of ankle strategy in the prosthetic limb. The ankle muscles are essential in recovering from a trip (Pijnappels et al., 2004) and are the primary contributors to controlling whole-body sagittal plane angular momentum (Neptune and McGowan, 2011).

Less able amputees will profit from a prosthetic foot with a large radius of curvature to improve standing stability (*Chapter 2;* Curtze et al., 2009). In the present study, the amputees wore their habitual prosthesis equipped with a variety of different prosthetic feet. While the general response patterns described in this article were strong, this variation in prosthetic feet may have had an impact on an individual's response to the perturbations. In future research, it would be interesting to systematically manipulate the radius of curvature of the prosthetic foot and test its effect on an individual's balance response.

To recover from perturbations in the medio-lateral direction, amputees as well as controls used the load-unload mechanism, which is driven by modulation of the hip moments. In previous research this has been shown to be the common balance strategy, when standing feet side-by-side (Winter et al., 1996). Despite the loss of part of their lower limb, transtibial amputees have two intact, muscle empowered hip joints. This is why they experienced little limitations in this perturbation condition. Training of the abductor/adductor muscles in rehabilitation may be beneficial to counter muscle atrophy in the amputated limb, thereby increasing the capacity to respond to balance perturbations.

A limitation of the waist-pull method is its difference to being pushed by others in crowded environments, as in the waist-pull method the direction of perturbation is predictable. Furthermore, resisting a pulling force, which is suddenly ceased, thereby initiating a fall is mechanically different from being pushed. However, in both situations the CoM is accelerated towards the boundary of support. In other studies the waist-pull is applied as a tug from a falling load (Pai et al., 1998 Hsiao-Wecksler et al., 2003). The limitation of this procedure is that the participant has to recover balance while the dangling load attached to the waist. However, the waist-pull method allows investigating the response behaviour to small perturbations in a systematic fashion. In the present study, the balance control strategies were shown to depend on the direction of perturbation, i.e. perturbations in anterior-posterior direction were compensated by the ankle strategy, while in medio-lateral direction the load-unload mechanism was active.

For the simplicity of the model presented an equal height of the greater trochanter and the CoM are assumed. This approximation allows a well enough delimitation of the contribution of the joint moment integral of the hip and ankle to balance control. Refining the model to unequal height of the CoM and greater trochanter as well as including complex segment dynamics is possible; however this would be at the expense of the simplicity of the model.

A key finding was that the passive properties of the prosthetic foot contribute to balance control in anterior-posterior direction. As the contribution of a prosthetic foot varies between models, this should be considered in the prosthetic fitting process. To optimize the patient-prosthesis match, an amputee's control and muscle strength in the intact hip and ankle joints and the amount of stability provided by the prosthetic device (*Chapter 2;* Curtze et al., 2009; Nederhand et al., 2011) should complement each other, for instance by applying balance measures (*Chapter 4*; Curtze et al., 2010b).

Suppliers

- Advanced Mechanical Technology, Inc., 176 Waltham Street, Watertown, MA 02472-4800, USA.
- vicon Motion System, 14 Minus Business Park, West Way, Oxford, OX2O-JB, UK.
- Brosa AG, Dr. Klein Straße 1, 88069 Tettnang, Germany.
- The MathWorks, Inc., Crystal Glen Office Centre, 39555 Orchard Hill Place, Suite 280, Novi, MI 48375, USA

Conflict of Interest

The authors declare to have no conflict of interest in this work.

Acknowledgements

This work was supported by OIM Holding BV, The Netherlands and Stichting Beatrixoord Noord-Nederland, The Netherlands. Furthermore, the authors thank Ronald Davidsz for his assistance with the data collection.

Balance Recovery After an Evoked Forward Fall in Unilateral Transtibial Amputees

6

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Gait & Posture (2010) 32: 336–341

Abstract

Falls are a common and potentially dangerous event, especially in amputees. In this study, we compared the mechanisms of balance recovery of 17 unilateral transtibial amputees and 17 matched able-bodied controls after being released from a forward-inclined orientation of 10%. Kinematic analysis revealed statistically significant differences in response time and knee flexion at heel-strike between both groups. However, there were no statistically significant differences in step length of the leading and trailing limb, swing time of the leading limb, and maximal knee flexion during swing. In the amputees, we found spatial and temporal diff erences when recovering with the sound versus prosthetic limb first. When leading with the prosthetic limb, they responded faster and also the interval between heel-strike of the leading and trailing limb was shorter. Furthermore, amputees made a longer step and showed less knee flexion at heel-strike when leading with the prosthetic limb. Interestingly, amputees as a group had no specific limb preference, prosthetic or sound, to recover after a forward fall, despite the asymmetry in their locomotor system. Analyses of dynamic stability (extrapolated center of mass) revealed that the amputees were equally efficient in recovering from an impending fall as controls, irrespective whether they lead with their prosthetic or sound limb. We suggest that in amputee rehabilitation, balance recovery after a fall should be trained with both sides, as this can increase confidence in fall-prone situations.

Introduction

In a rapidly ageing society, falls are a particular issue of concern for public health. Some patient populations have an even higher incidence of falls. The annual fall incidence, of one or more falls, in lower limb amputees is approximately 50%, while their elderly able-bodied community-dwelling peers have an incidence of 30–40% (Miller et al., 2001a; Miller et al., 2001b). While falls in the elderly population have received considerable attention, little is known about the underlying mechanisms of falls and successful balance recovery in lower limb amputees.

A popular approach to study balance recovery after a forward fall without compromising safety is the tether-release method (Do et al., 1982; Hsiao-Wecksler, 2008): participants recover balance after being suddenly released from a forward-inclined orientation by stepping (Figure 6.1). A main focus

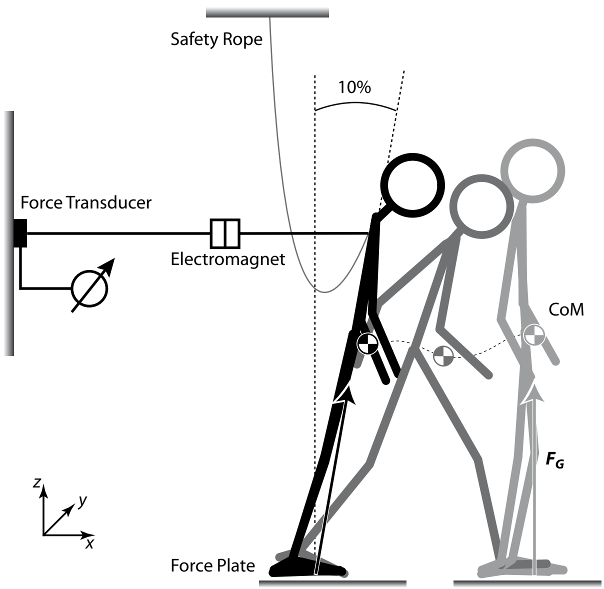

Figure 6.1 | Experimental setup. The participant was held in a forward-inclined orientation (10%). The force transducer was used to determine the magnitude of forward inclination which could be adjusted by changing the length of the cable. The participant was positioned on a force plate. A safety rope was attached to the full-body harness. The electromagnet was used as release mechanism, which could be triggered by a switch ceasing the current. After release the center of mass (CoM) gained velocity, a ground reaction force (***F_G***) acted on the CoM (real coordinates).

of previous studies has been the threshold between successful balance recovery and falls in young and elderly, by means of gradually increasing the initial forward-inclined orientation (Thelen et al., 1997; Wojcik et al., 1999); for a review see (Hsiao-Wecksler, 2008).

In the event of a forward fall, the center of mass (CoM) gains velocity, away from the base of support. The impending fall can only be prevented by a rapid and spatially well-directed stepping (or reaching) movement. The accurate, time dependent, step length can be predicted by the concept of the "extrapolated center of mass" (XcoM) (Arampatzis et al., 2008; Hof et al., 2005; Hof, 2008). This concept is based on the inverted pendulum model of balance and allows to define stability in dynamic situations (Hof et al., 2005; Hof, 2008). The purpose of this study is to provide an in depth analysis of the mechanisms of balance recovery in lower limb amputees. Furthermore, we hypothesized that unilateral transtibial amputees compensate for the asymmetry in their locomotor system by stepping with their sound limb first, as they lack active control of the ankle joint when stepping with their prosthetic limb first.

Methods

Participants

A group of 17 male unilateral transtibial amputees was included in the study. Their mean age was 55.2 (±9.2) years and they had a height of 1.84 (±0.07) m, and a body weight of 85.1 (±10.7) kg. The mean time since amputation was 12.6 (±13.7) years. Twelve participants had undergone an amputation of their left limb, and fi ve of their right limb. The most frequent reason of amputation was trauma (12), followed by vascular disease (4), and limb deficiency (1). The amputees were all experienced walkers, who used their prosthesis on a daily basis. The control group consisted of 17 healthy male participants matched in age (55.0 ± 10.3 years), height (1.85 ± 0.05 m) and weight (87.1 ± 9.1 kg).

Apparatus

The ground reaction forces during standing in a forward-inclined orientation were measured by an AMTI force plate, sampled at a rate of 1000 Hz (Figure

6.1). A second force plate in front of the participant was used in some of the trials only, when a participant happened to recover balance with both feet on this second force plate. One of these trials is presented in the results section for illustration purposes. To record full-body kinematics, 35 reflective markers were attached to the participant's anatomical bony landmarks as specified in the Vicon Plug-in Gait full-body model. On the prosthetic limb the markers were placed at the corresponding positions. The reflective markers were tracked by an eight-camera Vicon motion capture system at a sampling rate of 100 Hz. Anthropometric measurements were taken for each individual according to the Vicon requirements and fed into the model. The lean-control cable was equipped with an electromagnet as release mechanism. The participant's instantaneous lean angle was determined by means of a force transducer. Finally, all measurement data were further processed using Matlab.

Procedure

To evoke a forward fall, participants were suddenly released from a fixed forward-inclined orientation of 10% (≈6°; Figure 6.1). The participants were held back by a cable fixed to a full-body safety harness. The magnitude of the forward lean angle was controlled by adjusting the lean-control cable length until the force transducer attached to the cable indicated that it supported 10% of the participant's body weight. The forward fall was initiated by the experimenter by pressing a button releasing the electromagnet. The participants were verbally instructed to prevent themselves from falling. No specifications were given regarding a desired balance recovery technique. Participants were given one practice trial. After three trials they were requested to initiate the recovery by stepping with the other, non-preferred limb first. To prevent the participants from falling to the ground in the event of failed recovery, they were secured by a safety rope attached between the harness and the ceiling.

Prior to testing, all amputees completed the activities-specific balance confidence (ABC) scale (Miller et al., 2002; Miller et al., 2003; Powell and Myers, 1995), a self-efficacy measure assessing balance confidence across 16 specific activities on an analogue scale (0–100%).

The experimental protocol was approved by the local Medical Ethics Committee. All participants gave their informed consent.

Data analysis

We determined the leading limb preference (sound/prosthetic, left /right) from the first three trials of the fall experiment. Furthermore, trials were classified as single stepping, if the length of the trailing step did not exceed the length of the leading step by more than half a foot length. All other trials were classified as multiple stepping. The step length was normalized by the leg length resulting in a dimensionless number (Hof, 1996).

The following temporal characteristics were determined: (1) response time (ms), defined as the interval between release (t_0) and toe-off of the leading limb, (2) swing time (ms), defined as the interval between toe-off and heel-strike of the leading limb, and (3) heel-strike interval (ms), defined as the interval between heel-strike of the leading and trailing limb. The different events were estimated form the force transducer, the force plate data and the marker data of the heel and toe. Additionally, the maximal knee flexion of the leading limb during swing (°) and the knee flexion at heel-strike (°) were determined from the kinematic data.

The XcoM (Hof et al., 2005) was used to determine the function of the leading limb in braking the forward fall. Here, the forward position of the XcoM (ξ) is defined as

$$\xi = x_{CoM} + \frac{v_{xCoM}}{\omega_0}$$

in which x_{CoM} is the forward position of CoM, and v_{xCoM} the forward velocity oft he CoM. The eigenfrequency of a pendulum, a constant related to stature, is denoted as ω_0

$$\omega_0 = \sqrt{g/l}$$

where l is the effective pendulum length (trochanteric height times 1.24) (Winter, 1979). When the CoP is placed on the XcoM a stable posture will result. When the CoP is placed at some distance from the XcoM, the XcoM will move away from the CoP.

In a successful trial the CoP under the leading limb is positioned slightly beyond the XcoM. As a result the XcoM will start to move backward, the CoM movement will slow down and the fall will eventually be arrested. When balance will be successfully restored with the leading limb the XcoM thus shows a maximum, i.e. its first derivative is zero. Stability can be regained either with the leading limb, before heel-strike of the trailing limb, or after. Each trial was classified accordingly.

Statistical analysis

Pearson's chi-square test was used to control for differences in leading limb preference in the group of amputees (prosthetic, sound) and controls (left, right). Subsequently, a one-way ANOVA was performed to test if amputees with prosthetic and sound limb leading preference differed in balance confidence.

For each leading limb condition the mean values of the spatial and temporal characteristics were calculated. Separate two-way ANOVAs for repeated measures with leading limb (prosthetic/left, sound/right) as within-subject factor leading limb preference (prosthetic/left, sound/right) as between-subject factor were run on the different outcome parameters, first for the group of amputees and controls individually, and then over all participants. Furthermore, the relation of stepping strategy (single, multiple) and leading step length was determined (Pearson correlation).

Normality of data distribution (Kolmogorov–Smirnov test) was not given for one outcome measure, the percentage of trials in which balance was recovered with the leading limb. Here, non-parametric testing (Wilcoxon signed-rank test, Mann–Whitney test) was applied.

The level of significance was set to $p \leq .05$. All statistical analyses were performed using SPSS 16.0.

Results

All participants, amputees and healthy controls, were able to recover balance with any of both limbs leading after being released from a forward-inclined orientation of 10%. A failed recovery occurred only in one amputee participant who slipped on the floor after release when leading with his sound limb. No statistically significant differences in leading limb preference were found for the amputees (41.2% prosthetic, 58.8% sound) and the controls (52.9% left, 47.1% right) ($\chi^2(1, N = 34) = .119, p = .73$). The group of amputees with prosthetic limb leading preference as well as the sound limb leading preference group scored high on the ABC score (92.9 ± 7.2 and 89.5 ± 11.3 respectively). No statistically significant differences in balance confidence were found between these two amputee groups ($F(1,15) = .461, p = .508$).

Finally, we analyzed the contribution of the leading limb in arresting the forward fall. Figure 6.2 shows a single step recovery of a participant. While leaning in a forward-inclined orientation the CoP was posterior to the CoM, and the participant was just being held back by the lean-control cable. Upon release the CoM started to gain velocity, i.e. the XcoM and the CoM moved apart. By stepping the participant brought his base of support over the XcoM, thereby breaking the forward fall. Shortly after heel-strike of the leading limb, the forward motion of the XcoM reached its maximum. In this study, this was taken as a marker that balance starts to be recovered even before heel-strike of the trailing limb. After heel-strike of the trailing limb, the participant was standing upright and CoM, XcoM and CoP coincided. However, this illustration also reveals some minor inaccuracies. The concept of the XcoM is based on the inverted pendulum model, simplifying the human body as a pendulum with a mass, but recovering from a forward fall also involves accelerations of the swing leg. In theory the CoP should be positioned beyond the XcoM in order to move it back, while in Figure 6.2 the CoP is only close to the XcoM.

The percentage of trials in which balance was recovered with the leading limb is given in Figure 6.3. The group of amputees, who prefer leading with their prosthetic limb, recovered balance in a single step, when leading with their preferred prosthetic limb in 53.6% of the trials and when leading with their non-preferred sound limb in 42.9% of the trials. Interestingly, amputees who had a sound limb leading preference recovered balance in the leading step more oft en when leading with their non-preferred prosthetic limb (78.0%), then when leading with their preferred sound limb (65.0%). Non-parametric testing revealed no statistically significant differences in balance recovery for leading with the prosthetic versus sound limb in amputees ($z = -.89, p = .374$), and left versus right limb in controls ($z = -.21, p = .833$). Furthermore, the

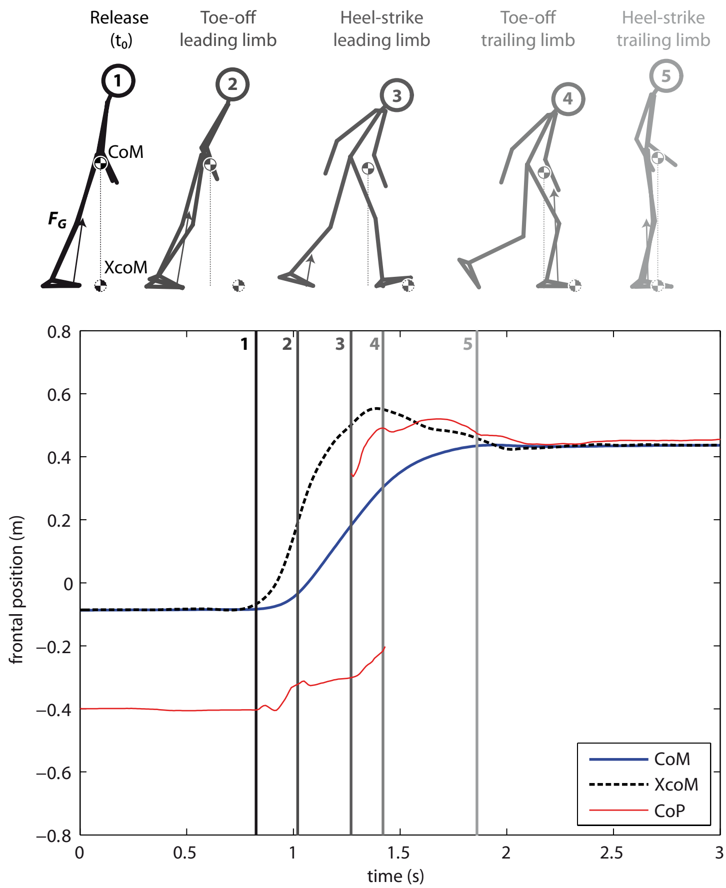

Figure 6.2 | Forward position of the CoM, XcoM, and CoP as a function of time. The (control) participant was released from a forward-inclined orientation and recovered balance with a single step. Sequence of events: (1) release at t_0, (2) toe-off leading limb, (3) heel-strike leading limb, (4) toe-off trailing limb, and (5) heel-strike trailing limb.

leading limb preference did not have a statically significant effect on balance recovery in amputees ($U = 25.5$, $n_1 = 10$, $n_2 = 7$, $p = .346$), nor in controls ($U = 26.5$, $n_1 = 9$, $n_2 = 8$, $p = .352$). In addition, amputees and controls did not differ on the extent to which they recovered balance with the leading limb ($U = 117$, $n_1 = 17$, $n_2 = 17$, $p = .336$).

In depth comparative analysis of the mechanisms of balance recovery revealed the following mean values for the special and temporal parameters given in Table 6.1. In lower limb amputees the step length of the leading limb

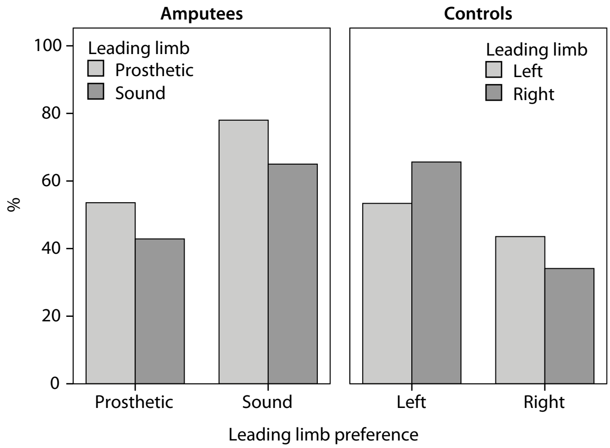

Figure 6.3 | Balance recovery with leading limb. Percentage of trials in which balance was recovered with the leading limb, i.e. the first derivative of XcoM was zero before heel-strike of the trailing limb (see Figure 6.2).

was statistically significant longer when leading with the prosthetic limb ($F(1,15) = 9.71$, $p < .001$), while no such step length asymmetries were found in controls ($F(1,15) = .28$, $p = .606$). Besides, there was no statistically significant correlation between stepping strategy, multiple or single, and the leading step length ($r = -.085$, $p = .492$).

In amputees, the step length of the trailing limb was statistically significant longer, when leading with the prosthetic and trailing with the sound limb ($F(1,15) = 4.86$, $p = .043$). Moreover, amputees, who had a preference for leading with their sound limb, made longer steps with the trailing limb, irrespective of the leading limb condition ($F(1,15) = 13.25$, $p = .002$). In the control group, however, the step length of the trailing limb was not affected by their leading limb preference ($F(1,15) = .213$, $p = .651$), nor the leading limb condition ($F(1,15) = 1.71$, $p = .212$).

With respect to the temporal dynamics of balance recovery, amputees showed longer response times when leading with the sound limb ($F(1,15) = 14.03$, $p = .002$). Furthermore, leading limb preference had a statistically significant effect on response time, amputees with a prosthetic limb leading preference responded faster ($F(1,15) = 5.89$, $p = .028$). In the group of controls, response time was neither affected by the leading limb

Table 6.1 | Spatial and temporal characteristics.

Group	**Amputees (n = 17)**					**Controls (n = 17)**				
Preference	**Prosthetic (n =7)**		**Sound (n = 10)**			**Left (n = 9)**		**Right (n = 8)**		
Leading	**Prosthetic**	**Sound**	**Prosthetic**	**Sound**		**Left**	**Right**	**Left**	**Right**	
Step length (leading)	0.877 (0.124)	0.815 (0.154)	0.857 (0.116)	0.721 (0.141)	l***	0.821 (0.104)	0.843 (0.096)	0.741 (0.092)	0.737 (0.125)	
Step length (trailing)	0.959 (0.184)	0.771 (0.137)	1.247 (0.136)	1.093 (0.413)	p**, l*	1.038 (0.330)	1.131 (0.360)	1.157 (0.363)	1.169 (0.331)	
Response time (ms)	196 (22)	220 (19)	216 (18)	246 (35)	p*, l**	238 (40)	243 (10)	239 (42)	236 (31)	g*
Swing time (ms)	367 (87)	208 (32)	341 (38)	303 (48)		301 (26)	303 (28)	297 (52)	312 (60)	
Heel-strike interval (ms)	591 (77)	680 (118)	503 (229)	741 (149)	l**	615 (215)	572 (127)	539 (111)	493 (124)	
Maximal knee flexion during swing (°)	64.1 (21.0)	62.6 (13.4)	61.7 (6.9)	66.2 (7.3)		65.9 (10.0)	64.8 (11.1)	61.7 (10.2)	59.7 (10.1)	
Knee flexion at heel-strike (°)	8.7 (7.5)	15.1 (5.3)	12.2 (4.7)	20.5 (7.6)	p*, l**	20.2 (6.4)	17.2 (7.4)	19.7 (8.7)	20.1 (7.4)	g**

mean (S.D.)

g, statistically significant group difference between amputees and controls

p, statistically significant between-group difference for leading limb preference (prosthetic/left, sound/right)

l, statistically significant within-subject difference for leading limb (prosthetic/left, sound/right)

* $p < .05$; ** $p < .01$; *** $p < .001$

Step length normalized for leg length.

condition ($F(1,15) = .01$, $p = .907$), nor by the leading limb preference ($F(1,15) = .02$, $p = .891$). Comparison between groups revealed that amputees responded statistically significant faster than controls ($F(1,32) = 4.53$, $p = .041$).

Analyses of the swing time in amputees yielded no marked differences between leading with the sound or prosthetic limb ($F(1,15) = 2.61$, $p = .127$). In addition, no statistically significant differences in swing time were found between amputees who preferred leading with their sound limb and those who preferred leading with their prosthetic limb ($F(1,15) = .26$, $p = .617$). Just like for the group of amputees, no effects of leading limb condition ($F(1,15) = 2.81$, $p = .114$) and leading limb preference ($F(1,15) = .02$, $p = .898$) on swing time were found for the group of controls.

In amputees, the heel-strike interval was shortened when leading with the prosthetic limb ($F(1,15) = 10.84$, $p = .005$). However, the leading limb preference, prosthetic versus sound, did not have a statistically significant effect on the heel-strike interval ($F(1,15) = .05$, $p = .835$). Again, no statistically significant differences on the heel-strike interval for leading with the left or right limb ($F(1,15) = 2.63$, $p = .127$) and leading limb preference ($F(1,15) = 1.13$, $p = .307$) were found in controls.

Analyses of the kinematics of the maximal knee flexion during swing in amputees revealed no statistically significant differences for leading with the prosthetic or sound limb, which agrees with the intact knee function in the prosthetic limb ($F(1,15) = .26$, $p = .620$). In addition, the leading limb preference had no statistically significant effect on the maximal knee flexion during swing ($F(1,15) = .01$, $p = .928$). Analyses of the maximal knee flexion during swing in the control group did not reveal a statistically significant difference for leading with the left or right limb ($F(1,15) = .92$, $p = .353$). Furthermore, the maximal knee flexion during swing was not statistically significant different between controls who had a leading preference for the left or right limb ($F(1,15) = .94$, $p = .347$).

At heel-strike amputees showed a smaller knee flexion when leading with the prosthetic limb ($F(1,15) = 9.08$, $p = .009$). Moreover, amputees with a preference of stepping with the prosthetic limb first, had a smaller knee flexion at heel-strike ($F(1,15) = 4.81$, $p = .044$). The controls did no show statistically significant differences when leading with their right or left limb ($F(1,15) = .44$,

p = .517) or having a preference to step with the left or right limb first ($F(1,15) = .15$, $p = .704$). Comparison between both experimental groups revealed, that amputees had an overall smaller knee flexion at heel-strike than controls ($F(1,32) = 8.83$, $p = .006$).

Discussion

In this study we investigated the effect of lower limb amputation on balance recovery after an evoked forward fall. Despite the asymmetry of their locomotor system the amputees as a group had no specific preference for recovering with their prosthetic or sound limb. Interestingly, about 40% of the amputees had a preference to initiate recovery with their prosthetic limb. This is even more striking, considering that the fall itself was not unexpected, but only the moment of release. This means that they could plan their stepping response beforehand. In post-experimental interviews most participants indicated that the amputation did not affect their leading limb preference and stepping with the non-preferred limb first would feel unnatural. In the present set-up, amputees were just as capable to recover balance with their sound as their prosthetic limb; making a change in leading limb preference unnecessary. Furthermore, we found that amputees and controls were equally efficient in regaining stability with the leading limb, after being released from a forward-inclined orientation of 10%. In about 60% of the trials, amputees regained/initiated stability with the leading limb. But, as we were interested in studying the natural recovery behavior, participants were not limited in the number of steps taken or instructed to regain balance with the leading step already. This however implies that we may have underestimated their actual ability to recover balance with a single step.

Kinematical analysis of balance recovery in amputees revealed less knee flexion at heel-strike when leading with the prosthetic limb. This reduced knee flexion may have been associated with the larger step length of the leading limb. When stepping with the prosthetic limb first, amputees needed to take a longer step, as they were not able to actively shift the CoP forward under their prosthetic foot after heel-strike. A correction of the CoP position after heel-strike is only possible with active ankle control.

In order to recover balance with the leading step the CoP under the leading limb and the XcoM need to coincide, bringing the base of CoP under the position of the CoM is not sufficient, because the CoM would still have velocity and move away from the CoP.

When leading with the prosthetic limb, amputees shortened the heel-strike interval, hereby increasing stability. One of the amputees even performed jump like movements, nearly landing in synchrony with both limbs. To further minimize the impact forces acting on the stump, he bent his knee and quickly lowered his arms, thereby the vertical trunk acceleration was reduced and the $\boldsymbol{F_G}$ became smaller.

With respect to movement planning (Do et al., 1982) have suggested an invariant preparation phase when recovering balance. However, our results did not support this idea. In our study amputees responded faster when leading with their prosthetic limb, irrespective of their leading limb preference. Moreover, no interaction between leading limb preference and leading limb condition were found.

Previous studies have addressed the effect of instructions limiting the number of steps on the kinematics and kinetics of balance recovery (Cyr and Smeesters, 2007; Cyr and Smeesters, 2009). They showed that the leading step was nearly identical, irrespective of the stepping strategy. In this study we also did not find a systematic statistically significant effect of stepping strategy, single versus multiple, on the different outcome parameters.

A limitation of the tether-release method is its difference to stumbling in daily life, where tripping usually comes as a surprise. Furthermore, stumbling due to unplanned foot contact involves more complex dynamics. However, the observed behavior appeared to be surprisingly natural.

It is remarkable that our experienced amputee participants performed quite well in this challenging test, even with their prosthetic limb leading. Several of them experienced this as a pleasant surprise, which added to their confidence. This experience suggests that amputees may benefit from bilateral fall training in rehabilitation.

Suppliers

- Advanced Mechanical Technology, Inc., 176 Waltham Street, Watertown, MA 02472-4800, USA
- Vicon Motion System, 14 Minus Business Park, West Way, Oxford, OX-2OJB, UK
- Brosa AG, Dr. Klein Straße 1, 88069 Tettnang, Germany
- The MathWorks, Inc., 3 Apple Hill Drive, Natick, MA 01760-2098, USA
- SPSS Inc., 233 S. Wacker Drive, 11th floor, Chicago, IL 60606-6307, USA

Conflict of interest

The authors declare to have no conflict of interest in this work.

Acknowledgements

This work was supported by OIM Holding BV, The Netherlands and Stichting Beatrixoord Noord-Nederland, The Netherlands. Furthermore, the authors thank Ronald Davidsz for his assistance with the data collection.

Over Rough and Smooth: Amputee Gait on an Irregular Surface

7

Carolin Curtze, Bert Otten, At L. Hof & Klaas Postema

Gait & Posture (2011) 33: 292–296

Abstract

When negotiating irregular surfaces, the control of dynamic stability is challenged. In this study, we compared the adjustments in stepping behaviour and arm-swing of 18 unilateral transtibial amputees and 17 able-bodied participants when walking on flat and irregular surfaces. Experimental findings revealed that unilateral transtibial amputees reduced their gait velocity only slightly when walking on irregular surfaces. Analyses of the temporal gait characteristics, i.e. stride time, stance time, double-support time and step frequency, showed no statistically significant adjustments. Interestingly, the amputees did not increase the stability margins for lateral balance which were calculated based on the concept of the "extrapolated center of mass". Furthermore, they did not increase their step width, which was already wide when walking on the flat surface. However, amputees did increase the lateral component of relative arm-swing velocity in order to walk stable on irregular surfaces.

Introduction

When walking around in the real world, as opposed to an uncluttered laboratory, stability is challenged by environmental factors, such as uneven terrain. In this study the dynamic stability of walking is challenged by irregularities of the walking surface (Thies et al., 2005). This irregular surface induces small unpredictable changes in the way the limb is supported, forcing the walking person to adjust his/her gait pattern. During walking, the vertical projection of the center of mass (CoM) describes a sinusoidal trajectory, always remaining within the lateral boundaries described by the alternating step positions of the two feet. To define stability in such dynamic situations as walking, the velocity of the CoM needs to be accounted for. The "extrapolated center of mass" (XcoM; Hof et al., 2005) is equivalent to the position of the CoM plus its velocity divided by the pendulum eigenfrequency, a constant related to stature. The XcoM precedes the sinusoidal trajectory of the CoM. In order to achieve a stable gait, humans follow the simple rule of placing their feet at a certain distance behind and outward of the XcoM, thereby redirecting the movement of the XcoM and CoM. The distance at which the center of pressure (CoP) under the foot is placed to the XcoM at the time of foot contact is a measure for the stability of gait. Transfemoral amputees have been shown to walk with a greater margin between the average CoP position and the XcoM, particularly on the prosthetic side (Hof et al., 2007). This is to account for the potentially less accurate foot placement and lack of ankle moment control.

The purpose of the present study is an in depth analysis of the adaptations in leg and arm movements when walking on irregular surfaces. We predicted that unilateral lower limb amputees increase their margins of stability in order to walk stable on irregular surfaces. Moreover, an increase in the lateral component of arm-swing was hypothesized to be a compensatory mechanism (Otten, 1999).

Methods

Participants

Eighteen male transtibial amputees with a mean age of 55.6 ± 9.5 years, a height of 1.83 ± 0.05 m, and a body weight of 90.3 ± 14.37 kg (including prosthesis) were included in this study. The mean time since amputation was 11.7 ± 13.5 years. Lower limb amputations were either due to trauma (11), vascular disease (6) or limb deficiency (1). Thirteen amputees had undergone an amputation of their left limb and five of the right limb. All amputees were experienced walkers, wearing their prosthesis on a daily basis. None of the patients reported to have experienced falls over the past year. An overview of the individual patient characteristics is given in Table 7.1. The control group consists of 17 healthy male participants, with a mean age of 55.0 ± 10.3 years, a height of 1.85 ± 0.05 m and a weight of 87.1 ± 9.1 kg.

Apparatus

The custom made irregular surface (Figure 7.1) consists of randomly arranged triangular wooden prisms (a short rod with triangular cross section), placed under a strip of 3 mm thick carpet (Thies et al., 2005). The irregular surface

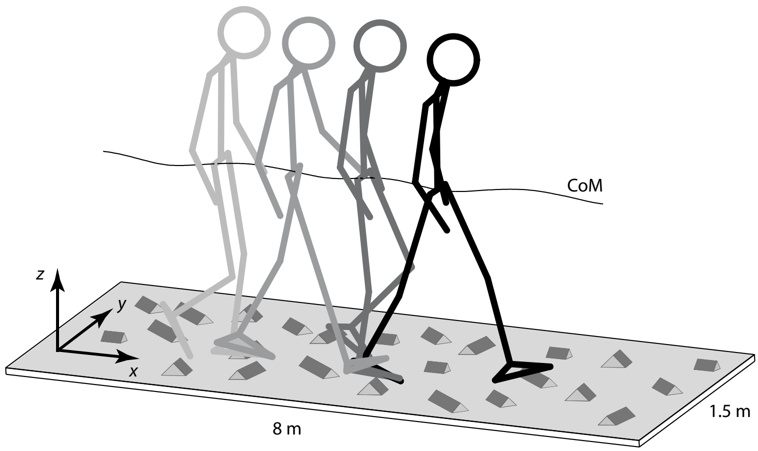

Figure 7.1 | Custom made irregular surface.

was 8 m long and 1.5 m wide. The density of wooden prisms was 26 prisms per square meter; each prism had a triangle base length of 30 mm, by a triangle height of 15 mm, and the prism length ranged between 60 and 160 mm.

To record full-body kinematics, 35 reflective markers were attached to the participant's anatomical bony landmarks as specified in the Vicon Plug-In-Gait full-body model. On the prosthetic limb the markers were placed at the corresponding positions. The reflective markers were tracked by an eight-camera Vicon motion capture system at a sampling rate of 100 Hz. Anthropometric measurements were taken for each individual according to the Vicon requirements and fed into the model. Finally, all measurement data were further processed using Matlab.

Table 7.1 | Participant characteristics.

	Height (m)	**Weight (kg)**	**Age (yr)**	**Time Since Amputation (yr)**	**Side of Amputation**	**Cause of Amputation**	**Prosthesis Components**
1	1.73	78.0	44	10	R	trauma	1C40 (Otto Bock)
2	1.79	98.0	53	44	L	trauma	1D10 (Otto Bock)
3	1.79	87.0	68	43	L	trauma	1C40 (Otto Bock)
4	1.79	63.0	50	16	L	trauma	1A30 (Otto Bock)
5	1.79	93.5	43	5	R	trauma	1A30 (Otto Bock)
6	1.80	83.0	51	7	L	trauma	1C40 (Otto Bock)
7	1.82	107.0	62	5	L	trauma	1C40 (Otto Bock)
8	1.89	94.5	53	4	L	trauma	1D35 (Otto Bock)
9	1.89	72.0	47	5	L	trauma	1C40 (Otto Bock)
10	1.90	91.5	34	6	L	trauma	1C40 (Otto Bock)
11	1.94	98.0	53	32	L	trauma	1C40 (Otto Bock)
12	1.81	83.0	66	7	L	vascular	1C40 (Otto Bock)
13	1.82	79.0	66	6	L	vascular	1D35 (Otto Bock)
14	1.82	117.0	61	4	L	vascular	Multiflex (Endolite)
15	1.82	85.5	64	2	R	vascular	1C40 (Otto Bock)
16	1.85	119.5	65	8	R	vascular	1D10 (Otto Bock)
17	1.87	89.0	60	2	R	infection	1D35 (Otto Bock)
18	1.87	86.5	60	4	L	limb deficiency	1D35 (Otto Bock)

Procedure

To simulate walking in a natural environment, the participants walked over an irregular surface which could be rolled out in the laboratory. In the control condition, participants walked on the flat walkway of the laboratory. Participants performed four trials for each of the two walking conditions. They were instructed to walk at their self-selected comfortable gait velocity. For reasons of safety, all participants wore a full-body safety harness suspended from an overhead track, inducing no forces on the participant.

Data analysis

To obtain insight into the dynamic stability of amputee gait on irregular surfaces, the lateral balance margins were determined based on the position of XcoM with respect to the supporting limb. The XcoM was computed based on the inverted pendulum model (Hof et al., 2005). The lateral position of the XcoM (ζ) was defined as

$$\zeta = y_{CoM} + \frac{v_{yCoM}}{\omega_0}$$

in which y_{CoM} denotes the lateral position of CoM, and v_{yCoM} the lateral velocity of CoM. Based on the effective pendulum length, trochanteric height times 1.34 (Massen and Kodde, 1979), the eigenfrequency was determined from

$$\omega_0 = \sqrt{g/l}.$$

Next, the margin of stability (b_{min}) was assessed as the minimal lateral distance between the anterior–posterior axis of the foot and the XcoM at foot contact. The anterior–posterior axis of the supporting foot was used as an approximation of the course of the CoP. The anterior–posterior axis was defined by the markers on the second metatarsal head and the calcaneus. As the participants drifted slightly in the horizontal plane, the trajectory of XcoM was detrended before calculating b_{min}.

Assessment of the spatio-temporal gait parameters included the gait velocity (m/s), the step length (m) and the step width (m). Furthermore, the stance time (from heel-contact to toe-off) as percentage of stride time and the double-support time (from heel-contact to opposite toe-off) as percentage of stride time were calculated for both limbs. In addition, the step frequency in each of the walking conditions was measured.

The lateral component of arm-swing was defined as the displacement of the wrist with respect to the shoulder joint center in lateral direction. The maximal velocity (m/s) of this component was reported.

Statistical analysis

For each of the walking conditions the mean values of the spatial and temporal walking characteristics were calculated. Separate ANOVAs for repeated measures with walking condition (flat, irregular surface) and limb (prosthetic/sound, left/right) as within-subject factor, and group (amputees, controls) as between-subject factor were run on the different outcome parameters, first over all participants, and then for each of the subgroups of amputees and controls. The level of statistical significance was set to $p \leq .05$. All statistical analyses were performed using SPSS 16.0.

Results

During walking, the vertical projection of the CoM describes a smooth sinusoidal trajectory, while the trajectory of the XcoM is more brisk (Figure 7.2A). The alternating stepping feet are placed behind and lateral to the XcoM. The lateral distance at which the foot is placed to the current XcoM position is a measure for the lateral stability (b_{min}). At every step the XcoM is redirected, turning sharply towards the contralateral side (Figure 7.2B).

Mean values of the stability margin bmin are given in Table 7.2. In amputees, this stability margin was not affected by the walking surface ($F(1,17) = .016$, $p = .690$). Furthermore, there was no statistically significant difference between the prosthetic and sound limb ($F(1,17) = 1.273$, $p = .275$). In the control group no statistically significant effect of walking surface on the stability margins were found ($F(1,16) = .004$, $p = .948$), nor a difference

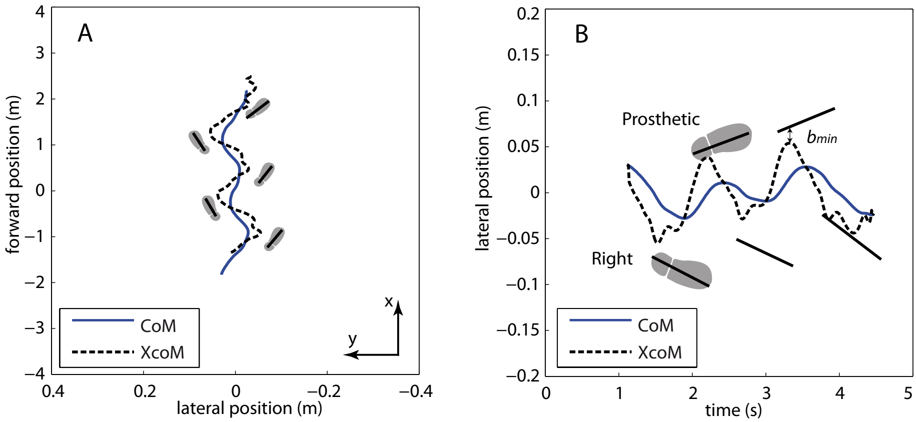

Figure 7.2 | (A) Trajectory of CoM and XcoM. Alternating step positions of the two feet and their anterior–posterior axis. Please note the difference in axis scaling of factor 10. Because of the difference in axis scaling the outward pointing of feet seems strongly exaggerated. **(B)** Lateral position of CoM and XcoM versus time. The margin of stability b_{min} is defined as the minimal distance between the anterior–posterior axis of the foot and XcoM at foot contact.

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between stepping with the left and right limb ($F(1,16) = .656$, $p = .430$). Comparison between the groups revealed no significant difference with respect to the stability margins ($F(1,33) = .092$, $p = .763$).

When walking on irregular surface gait velocity decreased significantly ($F(1,33) = 5.206$, $p = .029$). Furthermore, there were no significant differences in gait velocity between the two groups ($F(1,33) = 2.966$, $p = .094$). On subgroup level the decrease in gait velocity was not significant, neither in the group of amputees ($F(1,17) = 2.719$, $p = .118$), nor in the group of controls ($F(1,16) = 2.511$, $p = .113$).

Step length was significantly shorter when walking on the irregular surface ($F(1,33) = 6.916$, $p = .013$). The two groups differed with respect to their step length ($F(1,33) = 4.139$, $p = .050$). Analysis of the effect of walking surface on step length on subgroup level did not reveal significant effects for the amputees and the controls ($F(1,17) = 4.317$, $p = .053$ and $F(1,16) = 2.684$, $p = .121$, respectively).

No significant increase in step width was found for the study group as a whole ($F(1,33) = 4.041$, $p = .053$),however, the two groups differed significantly ($F(1,33) = 12.072$, $p < .001$). While there was no increase in step width in the amputee group ($F(1,17) = .001$, $p = .980$), controls increased their step width significantly ($F(1,16) = 23.278$, $p < .001$) when walking on the irregular surface.

With respect to temporal characteristics, we found no significant changes in stride time ($F(1,33) = 2.013$, $p = .165$) as an effect of walking surface. In addition, comparison of stride time between groups did reveal no significant differences ($F(1,33) = .733$, $p = .398$).

The walking surface did not have a significant effect on stance time as a percentage of stride ($F(1,33) = 1.195$, $p = .282$). Moreover, no significant group differences were found ($F(1,33) = .045$, $p = .833$). Analyses on group level revealed, that stance time as percentage of stride on the sound limb was significantly prolonged in amputees ($F(1,17) = 4.656$, $p = .046$). No such asymmetries in stance time were found in the group of controls ($F(1,16) = .015$, $p = .904$).

The double-support time as percentage of stride did not change as an effect of walking surface ($F(1,33) = 1.427$, $p = .241$). Furthermore, no significant differences in double support time were identified between amputees and controls ($F(1,33) = .031$, $p = .863$).

The walking surfaces did not have a significant effect on step frequency ($F(1,33) = 1.983$, $p = .168$). Both groups did not differ significantly with respect to their step frequency ($F(1,33) = .627$, $p = .418$).

Finally, the velocity of the lateral velocity of arm swing relative to the shoulder increased as an effect of walking surface irregularities ($F(1,33) = 4.460$, $p = .042$). The two groups differed significantly regarding arm-swing ($F(1,33) = 19.298$, $p < .001$). In the group of amputees a significant effect of walking surface was found ($F(1,17) = 5.924$, $p = .026$). While in the control group a statistically significant asymmetry in arm-swing was observed ($F(1,16) = 5.741$, $p = .029$), showing larger lateral velocities for the left arm.

Discussion

In the present study we analysed the dynamic stability of walking of unilateral transtibial amputees and able-bodied controls on irregular surfaces. Investigations of the foot placement with respect to XcoM revealed no changes in the lateral stability margin as an effect of waking surface. As visual feedback of surface irregularities was reduced by a strip of carpeting, anticipation of the exact direction of perturbation may have been impaired, making targeted

Table 7.2 | Spatial and temporal walking characteristics.

Group	Amputees (n = 18)				
Surface	**Flat**		**Irregular**		
Limb	**Prosthetic**	**Sound**	**Prosthetic**	**Sound**	**s, l**
b_{min} (mm)	17.6 (23.7)	24.6 (14.7)	15.9 (41.0)	21.9 (12.9)	
Gait velocity (m/s)	1.17 (.13)		1.12 (.21)		
Step length (m)	.68 (.07)	.70 (.07)	.66 (.10)	.67 (.10)	
Step width (mm)	108.5 (28.5)		108.6 (26.4)		
Stride time (s)	1.20 (.10)		1.21 (1.3)		
Stance time (% of stride)	65.5 (3.1)	67.7 (2.9)	65.9 (3.4)	67.7 (2.3)	l
Double-support time (% of stride)	15.6 (1.9)	17.4 (4.2)	15.6 (2.2)	17.8 (3.7)	
Step frequency (Hz)	1.70(.18)	1.68 (.15)	1.67 (.23)	1.68 (.17)	
Maximal lateral velocity of arm swing (m/s)	.274 (.091)	.260 (.070)	.303 (.125)	.293 (.104)	s

mean (S.D.)
G, statistically significant group difference between amputees and controls
S, statistically significant within-subject difference for walking surface for the group as a whole (amputees and controls)
s, statistically significant within-subject difference for walking surface on subgroup level (amputees or controls)
L, statistically significant within-subject difference for limb for the group as a whole (amputees and controls)
l, statistically significant within-subject difference for limb on subgroup level (amputees or controls)

adjustment in foot placement difficult. Furthermore, the margins for stepping with the sound or prosthetic limb did not differ; which is in contrast to findings of Hof et al. (2007). This divergence in findings may be due to the differences in level of amputation between study groups. Here, we investigated a group of unilateral transtibial amputees, while the study population of Hof et

Table 7.2 | *continued.*

Group	**Controls (*n* = 17)**					**Amputees & Controls (*N* = 35)**
Surface	**Flat**		**Irregular**			
Limb	**Left**	**Right**	**Left**	**Right**	**s, l**	**G, S, L**
b_{min} (mm)	16.0 (51.7)	19.2 (31.0)	13.6 (42.2)	23.2 (38.5)		
Gait velocity (m/s)	1.26 (.13)		1.21 (.17)			S
Step length (m)	.72 (.06)	.73 (.08)	.71 (.09)	.71 (.08)		G, S
Step width (mm)	71.3 (26.9)		83.8 (30.0)		s	G
Stride time (s)	1.17 (.08)		1.18 (.09)			
Stance time (% of stride)	66.4 (2.8)	66.5 (2.1)	66.6 (2.7)	66.7 (2.3)		
Double-support time (% of stride)	16.3 (2.4)	16.3 (3.2)	16.7 (2.4)	16.5 (3.1)		
Step frequency (Hz)	1.74 (.19)	1.73 (.15)	1.71 (.20)	1.70 (.16)		
Maximal lateral velocity of arm swing (m/s)	.202 (.086)	.162 (.094)	.223 (.086)	.156 (.057)	l	G, S

al. (2007) was a group of unilateral transfemoral amputees. The higher level of amputation may have a stronger effect on the accuracy of foot placement, resulting in a larger stability margin on the prosthetic side. However both groups of amputees lack active ankle control, which is essential for the fine-tuning of the position of the CoP during foot contact. A limitation of the present study was that we could not measure the exact position of the CoP during walking on the irregular surface, but used the anterior–posterior axis of the foot as an approximation of CoP position.

Notably, amputees did not increase step width (11 cm) when walking on irregular surfaces, they were stepping wide on the flat surface already. Controls however, did increase their step width of 7 cm when walking on flat surface by about 1 cm. This suggests that amputees choose not to increase stability by means of increasing their step with when walking on irregular surface. The gait velocity and step length decreased only slightly, this is why these effects reached statistical significance only for the group as a whole, but not on sub-group level.

Interestingly, the unilateral transtibial amputees showed no major adaptations in the temporal gait characteristics when negotiating irregular surfaces. This is in contrast to findings of Kendell et al. (2010), who found prolonged stride times and absolute double support times in unilateral transtibial amputees walking over foam mats, which is however a completely different condition. The changes described by Kendell et al. (2010) may have been an effect of reduced gait velocity; unfortunately no data was reported with respect to this.

Finally, we found that the lateral velocity of arm swing played a vital role in the control of dynamic balance. Amputees increased the maximum velocity of the lateral component of arm-swing when walking on irregular surface, which helps them to bring the CoM back over the base of support (Otten, 1999). In the controls lateral velocity of arm swing was asymmetric, which is a common phenomenon reflecting the functional differences: frequency control and stabilization. Studies in healthy subjects have shown that asymmetric arm-swing is not related to handedness, nor to asymmetrical leg movements (Kuhtz-Buschbeck et al., 2008).

A limitation of this study was that no variability measures could be determined due to the restricted number of strides in some of the participants. Future studies should aim at the effect of surface irregularities on the variability of amputee gait. In addition, it may be interesting to study if the cause of amputation affects the movement strategy. It is known that patients with vascular disease have compromised lower limb somatosensation, which is linked with poor balance. This reduced somatosensation and poor balance may influence outcome and movement strategy, when walking on irregular surface.

In this experiment we found that unilateral transtibial amputees were able to maintain a considerable gait velocity when walking on irregular surfaces, while adjusting their stepping pattern only slightly. Furthermore, our findings highlight the importance of the lateral component of arm-swing for stable prosthetic walking on irregular surfaces.

Suppliers

- Vicon Motion System, 14 Minus Business Park, West Way, Oxford, OX-2OJB, UK
- The MathWorks, Inc., 3 Apple Hill Drive, Natick, MA 01760-2098, USA.
- SPSS Inc., 233 S. Wacker Drive, 11th floor, Chicago, IL 60606-6307, USA.

Conflict of interest statement

The authors declare to have no conflict of interest in this work.

Acknowledgements

This work was supported by OIM Holding BV, The Netherlands and Stichting Beatrixoord Noord-Nederland, The Netherlands. Furthermore, the authors thank Ronald Davidsz for his assistance with the data collection.

Effects of Lower Limb Amputation on the Mental Rotation of Feet

8

Carolin Curtze, Bert Otten & Klaas Postema
Experimental Brain Research (2010) 201: 527–534

Abstract

What happens to the mental representation of our body when the actual anatomy of our body changes? We asked 18 able-bodied controls, 18 patients with a lower limb amputation and a patient with rotationplasty to perform a laterality judgment task. They were shown illustrations of feet in different orientations which they had to classify as left or right limb. This laterality recognition task, originally introduced by Parsons (1987), is known to elicit implicit mental rotation of the subject's own body part. However, it can also be solved by mental transformation of the visual stimuli. Despite the anatomical changes in the body periphery of the amputees and of the rotationplasty patient, no differences in their ability to identify illustrations of their affected versus contralateral limb were found, while the group of able-bodied controls showed clear laterality effects. These findings are discussed in the context of various strategies for mental rotation versus the maintenance of an intact prototypical body structural description.

Introduction

Our brain maintains multiple representations of the human body. One of them is the "body structural description", a topological map of the human body in general (Corradi-Dell'Acqua et al., 2009; Schwoebel and Coslett, 2005; Buxbaum and Coslett, 2001; Sirigu et al., 1991; Pick, 1922). In addition, for the spatial organization of movements, our brain maintains a dynamic mental representation of our own body and its relation to the external world. This so-called "body schema" (Head and Holmes, 1911) is an on-line, real-time mental representation of one's own body.

Neuropsychological studies suggest that the mental simulation of movement is guided by our body schema just like real motor actions (de Lange et al., 2005; Parsons, 1994; Schwoebel et al., 2001). During the mental simulation of movements, the motor system is subliminally activated; only the actual execution of the motor act is suppressed. This parallelism between imagined and actual movements is evidenced by functional neuroimaging, showing large overlaps in brain activity between the two. The superior parietal cortex, the intraparietal sulcus, and the adjacent rostral-most part of the inferior parietal lobe were found to be highly active during mental rotation of body parts (Bonda et al., 1995; Parsons et al., 1995), just as during real movements. Further, chronometric Findings revealed that mental rotation is subject to the same anatomical constraints as actual motor behaviour (e.g. Parsons, 1994).

In the present study we examined how physical changes in lower limb amputees affect these phenomena. After amputation of a limb, no sensory input will reach the deafferented regions of the cortex. Peripheral input is vital for the formation and stability of the cortical sensorimotor map. Through movement, the cortical representations of limbs are expanded (e.g. Gaser and Schlaug, 2003), and consequently the loss of a limb results in shrinkage of these representations (e.g. Knecht et al., 1996). In spite of this cortical reorganisation, many amputee patients experience a phantom limb. Some amputees describe their phantom as paralysed, while others can move it voluntarily (Melzack, 1992; Ramachandran and Rogers-Ramachandran, 2000). Blakemore et al. (2002) suggest that the motor control system can counteract the discrepancies between predicted and actual consequences of the motor command by modifying the forward model towards a no-movement prediction.

To quantify the effect of lower limb amputation on the mental rotation of feet, we used the laterality recognition task, first introduced by (Parsons, 1987). In this task, illustrations of hands or feet are presented in different views and orientations in random sequence. In the present study, participants are asked to classify foot illustrations according to their laterality as rapidly and accurately as possible by pressing a key. Participants were instructed to place their feet under the table (out of sight) in a forward pointing position and not to move them throughout the experiment. This task has been shown to elicit implicit mental rotation of one's own physical counterpart, until it is aligned with the position of the stimulus illustration. The more the orientation of the stimulus matches the orientation of the biological counterpart, the faster and more accurately it is recognized. Response time and error rate vary as function of stimulus orientation, reflecting the anatomical constraints of actual leg movements. Alternatively, this laterality judgment task can be solved by referring to a prototypical body schema, the "body structural description" (Corradi-Dell'Acqua et al., 2009), or visual transformation of the stimuli.

In addition to applying this task to lower limb amputees, we also tested a patient with a far more dramatic change of lower limb functional anatomy. This patient had undergone a Van Nes rotationplasty, a rare surgical interven-

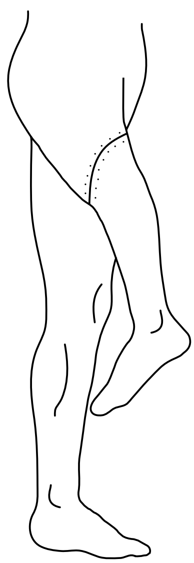


Figure 8.1 | Rotationplasty.

tion in which a diseased part of the femur bone and the knee is removed and the lower leg is re-attached to the thigh by a rotation through 180° (Figure 8.1; (Van Nes, 1950). The ankle now acts as a knee joint while the foot is facing backward. With this surgery the sciatic nerve is preserved. Further, patients do not experience phantom pain that is common in amputees (Grimer, 2005). In order to walk with a prosthesis, the motor control system is forced to adapt to the changed anatomical situation. Bending and stretching of the ankle joint result in altered movement effects. Hence, these adaptations should also be reflected in the mental transformation of feet, if the patient refers to her own body structure.

The goal of this study is to gain insight into the changes in body representation that take place due to lower limb surgery by means of a laterality judgment task that is assumed to require motor imagery (Bonda et al., 1995; Parsons et al., 1995). Determining the effects of absence of a limb (amputation) opposed to a case of transformation of a limb (rotationplasty) on mental rotation will advance our understanding of body representation.

For the group of amputees, a selective impairment of mental rotation for the absent limb was hypothesised, as was found for upper limb amputees (Nico et al., 2004). Furthermore, studies in healthy people have suggested that mental rotation depends on the actual posture of the physical counterpart of one's own body (de Lange et al., 2006; Ionta et al., 2007). If mental rotation is sensitive to these transformations, this should be reflected in the performance data of the patient with a rotationplasty.

Methods

Participants

Eighteen participants (14 male, 4 female; mean age 62.8 ± 9.7 years) with a lower limb amputation (7 right, 11 left; level of amputation: 9 transtibial, 9 transfemoral) participated in the study (Table 8.1). The most frequent reasons for amputation were vascular disease (8), followed by trauma (7) and cancer (3). The mean time since amputation was 21.7 ± 20.2 years. All participants stated that they occasionally experience a phantom limb and are able

Table 8.1 | Participant characteristics.

	Sex	Age (yr)	Handedness	Amputation			
				Side	Footedness	Level	Time (yr)
Amputees							
1	M	51	R	L	R	Transtibial	26
2	M	55	R	L	R	Transfemoral	4
3	F	65	R	L	R	Transfemoral	46
4	M	71	R	L	R	Transtibial	15
5	M	62	R	L	R	Transfemoral	1
6	M	60	R	L	R	Transfemoral	41
7	M	49	R	L	R	Transfemoral	7
8	M	68	R	L	R	Transtibial	7
9	M	68	R	L	L	Transfemoral	3
10	F	66	R	L	R	Transfemoral	9
11	M	49	R	L	R	Transfemoral	30
12	F	57	R	R	R	Transtibial	5
13	F	66	R	R	R	Transtibial	39
14	M	78	R	R	R	Transtibial	58
15	M	50	R	R	R	Transtibial	25
16	M	76	R	R	R	Transtibial	63
17	M	60	R	R	R	Transtibial	3
18	M	79	R	R	R	Transfemoral	8
Rotationplasty							
1	F	37	L	R		Rotationplasty	21

to perform voluntary movements with their phantom. Additionally, one female participant (37 years) with a rotationplasty (right leg, time since surgery 21 years) took part in the study. All participants were prosthetic users, fitted with a mechanical prosthesis. Further, a group of 18 healthy control participants (13 male, 5 female; mean age 57.0 ± 10.9 years) was recruited. The protocol was approved by the local ethical committee. All participants gave their informed consent.

Table 8.1 | *continued.*

	Amputation		**Phantom limb**		
	Cause	**Prosthetic use**	**General**	**During testing**	**Quality**
Amputees					
1	Cancer	Yes	Yes	No	-
2	Vascular	Yes	Yes	No	-
3	Trauma	Yes	Yes	No	-
4	Vascular	Yes	Yes	No	-
5	Vascular	Yes	Yes	No	-
6	Trauma	Yes	Yes	No	-
7	Vascular	Yes	Yes	No	-
8	Vascular	Yes	Yes	No	-
9	Vascular	Yes	Yes	No	-
10	Trauma	Yes	Yes	No	-
11	Trauma	Yes	Yes	No	-
12	Cancer	Yes	Yes	No	-
13	Cancer	Yes	Yes	Yes	Resting in normal orientation
14	Trauma	Yes	Yes	No	
15	Trauma	Yes	Yes	Occasionally	Rotational movements
16	Trauma	Yes	Yes	No	-
17	Vascular	Yes	Yes	No	-
18	Vascular	Yes	Yes	No	-
Rotationplasty					
1	Cancer	Yes	No	-	-

Patient description: rotationplasty

I.A. is a 37-year-old, left dominant, IT service assistant. She underwent a Van Ness rotationplasty of the right leg due to bone cancer at the age of 16. Since then she became a highly active prosthetic user. The patient reports that in the first weeks after the surgery she experienced touches on the left of her affected limb as being touched on the right, and vice versa. Also the perception of the dorsal and ventral part was inverted. In this early stage the patient was dependent on visual feedback for the recalibration of this tactile afferent information. Nights were particularly stressful; then she experienced her affected leg hovering in the air while the bed's full weight was pressing down her affected leg. After reorganisation, she re-referenced tactile afferent information; thus, she was no longer dependent on visual feedback. However, till date, as was specified by the patient, she has to reckon the position of her big toe on the affected side.

Stimuli

Stimuli in the laterality recognition task were the original line drawings by (Parsons, 1987) and matching photos of prosthetic and natural feet (Figure 8.2). The illustrations of left and right feet were mirror images. They depicted the dorsal and plantar view of feet in six different orientations (rotation angle, 60° steps). This results in a total of 72 different stimuli. The zero orientation was defined as toes pointing up for the dorsal view and toes pointing down for the plantar view.

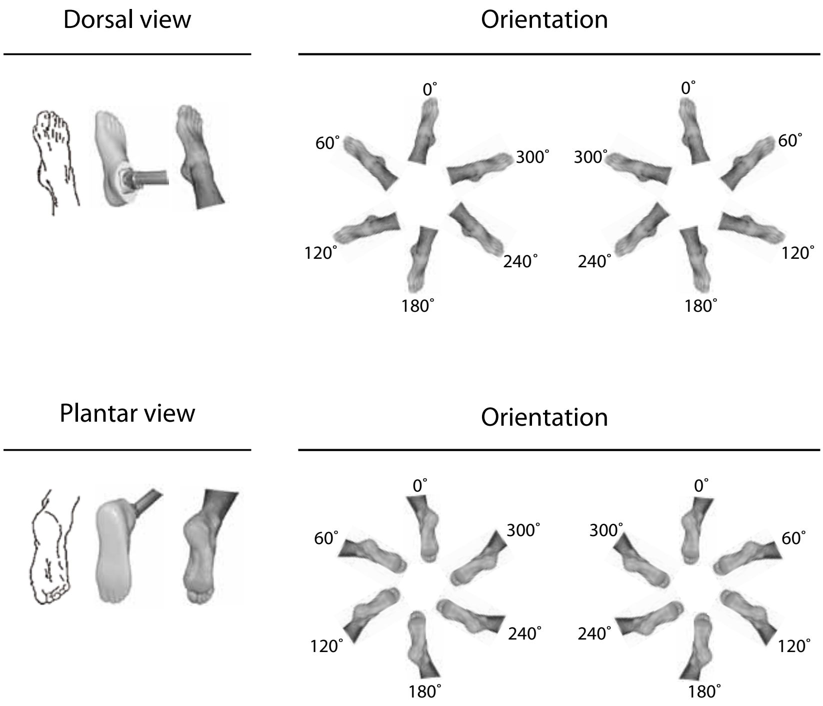


Figure 8.2 | Stimuli. Exemplary stimulus material of the right foot; the corresponding left foot stimuli were mirror images. Stimuli were presented in two views (dorsal, plantar) and six different orientations (60° steps).

Design and procedure

The study was designed as an experimental, cross-sectional study with a control group. Participants were required to judge foot illustrations that were presented as single images on a monitor for their laterality. The participants sat comfortably in front of the computer, with their left and right index finger placed on the two response keys. They positioned their feet parallel and out of sight under the table. They were instructed not to move their feet during the experiment. Preceding each stimulus, a fixation cross appeared in the middle of the blank screen and remained visible for 1 s. As it disappeared the stimulus was presented in the same location. The participants were instructed to respond as quickly and as correctly as possible by pressing the left or right key (corresponding to identification as left or right limb). If they did not respond within a time window of 5 s, the test proceeded to the next trial automatically. The sequence of stimuli was randomized for each participant.

The participants completed a practice phase, in which each of the 72 different stimuli was presented to them once. The subsequent experimental phase consisted of 4 blocks with 144 trials each.

Data analysis

Analyses included response time for accurate response trials only. Trials with response times below 500 ms and above 3500 ms were excluded from further analyses. To correct for skewness of the data distribution, response times were logarithmically transformed. The error rate was calculated and arcsine transformed.

Two separate three-way ANOVAs for repeated measures with stimulus laterality (left, right), view (plantar, dorsal) and orientation (0°, 60°, 120°, 180°, 240° and 300°) as within-subject factors were performed on the response time and the error rate of the controls.

To evaluate the effect of lower limb amputation, two separate four-way ANOVAs with stimulus laterality (affected, contralateral), view (plantar, dorsal), orientation (0°, 60°, 120°, 180°, 240°, 300°) as within-subject factors, and amputated limb (left, right) as between-subject factor were run on the response time and the error rate.

To determine the effect of the presence/absence of phantom sensations during the task two separate four-way ANOVAs were run on the response time and the error rate. The within-subject factors were stimulus laterality (affected, contralateral), view (plantar, dorsal), and orientation (0°, 60°, 120°, 180°, 240°, 300°); and phantom (presence/ absence of occasional phantom sensations during the task) was the between-subject factor.

Furthermore, to evaluate the effect of time since amputation, two separate analyses of covariance with stimulus laterality (affected, contralateral), view (plantar, dorsal), and orientation (0°, 60°, 120°, 180°, 240°, 300°) as within-subject factors and time since amputation as covariate were performed.

To compare the controls with the amputees, two separate four-way ANOVAs were run on the response time and the error rate. The within-subject factors were stimulus laterality (left/affected, right/ contralateral), view (plantar, dorsal), and orientation (0°, 60°, 120°, 180°, 240°, 300°); and group (controls, amputees) was the between-subject factor.

To further analyse the effect of lower limb surgery, ratios between affected and contralateral limb were calculated from the logarithmic transformed response time (RT) data ($\ln(RT_{affected}/RT_{contralateral})$) for the amputees and the patient with rotationplasty, as well as for the left and right limb of healthy controls.

Contrasting amputees and rotationplasty, two-tailed one-sample Z tests were used to determine if the observed logarithmically transformed ratios in response time are likely to come from different populations.

Results

Controls

The three-way ANOVA on the response time data indicated significant main effects of view ($F(1,17) = 115.50$, $p < .001$), and orientation ($F(5,85) = 137.38$, $p < .001$). Feet in the plantar view were considerably more difficult to recognize than feet in the dorsal view (Figue 8.3). Controls were also faster when recognizing feet presented in a more natural orientation. Moreover, the interaction between view × orientation reached statistical significance ($F(5,85) = 29.81$, $p < .001$), while the interaction between stimulus laterality × orientation was non-significant ($F(5,85) = 1.21$, $p < 0.309$). Furthermore, there was a significant main effect for stimulus laterality ($F(1,17) = 13.76$, $p = .002$), indicating faster mental rotation for right than left feet.

Analyses of the error rate (Figure 8.4) confirmed these significant main effects (view $F(1,17) = 14.15$, $p = .002$; orientation $F(5,85) = 7.63$, $p < .001$; stimulus laterality $F(1,17) = 1.39$, $p = .254$). The interaction view × orientation ($F(5,85) = 4.25$, $p = .002$), while the interaction of stimulus laterality × orientation remained non-significant ($F(5,85) = 0.99$, $p < .429$).

Amputees

For the amputees, we found significant main effects on response time for view ($F(1,16) = 202.92$, $p < .001$), and orientation ($F(5,80) = 74.54$, $p < .001$) as well as a significant interaction between these two (view × orientation $F(5,80) = 12.80$, $p < .001$), just as for healthy controls. Interestingly, no significant main effects were observed for either stimulus laterality ($F(1,16) = .62$, $p = .445$) or amputated limb ($F(1,16) = .01$, $p = .927$), nor the interaction between those two (stimulus laterality × amputated limb $F(1,16) = 3.18$, $p = .094$). Only the interaction of stimulus laterality × orientation ($F(5,80) = 2.41$, $p = .044$) reached some level of statistical significance. Remarkably, the presence/absence of a phantom limb during task execution did have a statistically significant effect on the laterally judgments (stimulus laterality × orientation × phantom $F(5,80) = 5.44$, $p < .001$). Time since amputation, however, did not have a statistically significant effect on task performance (stimulus laterality × time since amputation $F(1,16) = 2.93$, $p = .106$). Further, amputees who had lost their dominant limb did not perform any different from amputees who had lost their non-dominant limb (stimulus laterality × dominant/non-dominant limb loss $F(1,16) = 1.92$, $p = .185$).

Analysis of the error rate confirmed these findings (view $F(1,16) = 39.78$, $p < .001$; orientation $F(5,80) = 18.88$, $p < .001$; view × orientation $F(5,80) = 8.63$, $p < .001$; stimulus laterality $F(1,16) = .01$, $p = .911$; amputated limb $F(1,16) = .53$, $p = .478$; stimulus laterality × amputated limb $F(1,16) = 2.18$, $p = .159$; stimulus laterality × orientation $F(5,80) = 0.14$, $p = .984$).

Contrasting the group of amputees with the controls in two separate four-way ANOVAs, no significant main effect emerged for the factor group (re-

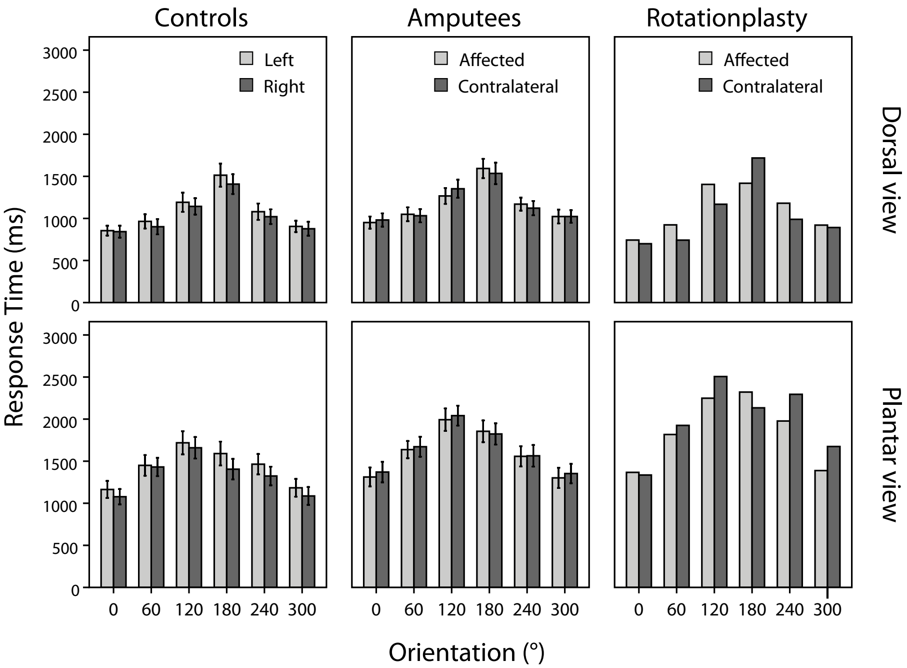


Figure 8.3 | Response times. Mean response times by group (controls (n = 18), amputees (n = 18) and rotationplasty (n = 1)) for different views (dorsal, plantar) and stimulus laterality (affected/left and contralateral/right) are shown. Error bars depict ± 1 S.D.

sponse time: $F(1,34) = 2.07$, $p = .159$; error rate: $F(1,34) = 1.37$, $p = .250$). We found a significant main effect for the factor view (response time: $F(1,34) = 323.12$, $p < .001$; error rate: $F(1,34) = 50.61$, $p < .001$). There was no significant interaction between view × group found for response time ($F(1,34) = 2.70$, $p = .110$), but only for error rate ($F(1,34) = 4.18$, $p = .049$). Additionally, the orientation of stimuli had a significant effect on task performance (response time: $F(5,170) = 201.38$, $p < .001$; error rate: $F(5,170) = 22.69$, $p < .001$), but there was no significant interaction between orientation × group (response time: $F(5,170) = .57$, $p = .726$; error rate: $F(5,170) = .209$, $p = .958$). Moreover, analyses revealed a significant main effect for stimulus laterality (response time: $F(1,34) = 4.86$, $p = .034$; error rate: $F(1,34) = .31$, $p = .580$). The interaction between stimulus laterality × group was significant only for response time ($F(1,34) = 7.62$, $p = 0.009$) but not for error rate ($F(1,34) = .22$, $p = .640$).

In a covariance analysis on all subjects (controls and amputees), we

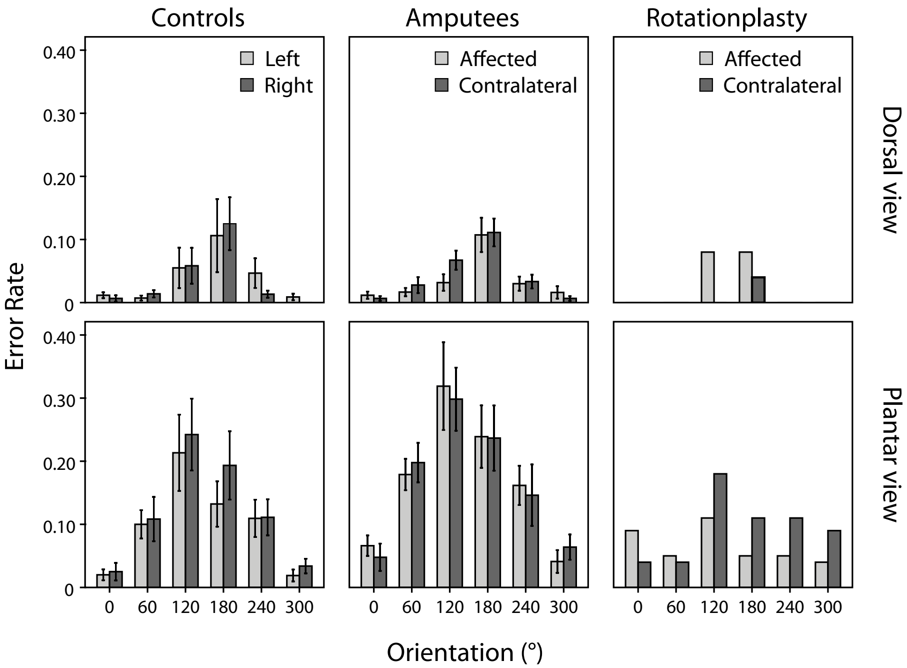


Figure 8.4 | Error rate. Mean error rate by group (controls (n = 18), amputees (n = 18) and rotationplasty (n = 1)) for different views (dorsal, plantar) and stimulus laterality (affected/left and contralateral/right) are shown. Error bars depict ± 1 S.D.

checked for a possible influence of motor preferences (handedness) on the results. This analysis showed no interaction for stimulus laterality × handedness ($F(1,35) = .017$, $p = .898$), so we conclude that the laterality effect is not due to differences in motor preferences.

Rotationplasty

Visual inspection of the response time data of the patient with rotationplasty (Figure 8.3) revealed the characteristic effects of orientation and view. Likewise, the error rate increased with the degree of orientations and the plantar view (Figue 8.4), just as for all other participants. Despite the reverse foot position of the affected limb, there was no apparent interaction between orientation × stimulus laterality. Merely a slight trend towards faster and more accurate responses for the affected limb for the plantar view, but not the dorsal view, can be observed.

In Figure 8.5, the ratios between affected and unaffected limb are shown for comparison between amputees and rotationplasty. As shown, the intervals of 2 × S.D. encompass the mean values of the rotationplasty patient, indicating that these ratios fit within the normal range of amputees, as well as controls. Two-tailed one-sample Z tests yielded no significant group differences on response times (dorsal: $p \geq .11$; plantar: $p \geq .29$).

Post hoc power analysis

The fact that we did not detect a main effect of stimulus laterality (affected, contralateral) for the amputee group could possibly be attributed to a lack of power. We performed a post hoc power analysis, assuming a significance level

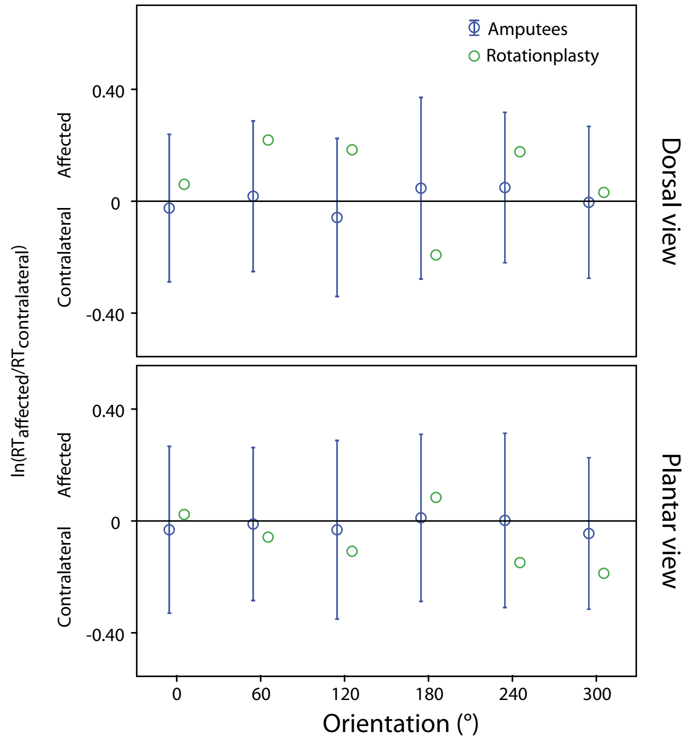


Figure 8.5 | Response time ratios. Mean ratios of response time (RT) of affected/left and contralateral/right limb for different views (dorsal, plantar) and orientations. Error bars depict ± 2 S.D. to indicate the significance level of the Z test (see Data analysis).

$\alpha = .05$. Given our sample size of 18 amputees, we should be able to detect medium effects ($\Delta = .25$) of stimulus laterality with high power ($1\text{-}\beta = .99$), and small effects ($\Delta = .10$) with moderate power ($1\text{-}\beta = .55$) (Cohen, 1988). Thus, it is unlikely that we would have missed an existing medium main effect of stimulus laterality due to lack of power.

Discussion

In the present study, we measured chronometric changes in the mental rotation of feet as a consequence of lower limb amputation and rotationplasty. Our data confirm the typically strong effects of view, orientation and stimulus laterality found in healthy controls (de Lange et al., 2006; Ionta et al., 2007), suggesting mental rotation of the subject's own physical counterpart into the position of the stimulus. The effect of stimulus laterality, i.e. right feet were recognized faster than left feet, is most likely to be explained by foot dominance (Parsons, 1987).

Our data on lower limb amputees indicate that these patients maintained their ability to perform mental rotations of feet representing the lost limb. Interestingly, no differences were observed in response time and error rate for judgments based on the affected versus contralateral limb. However, some differences in response were found between affected and contralateral limb, when accounting for the orientation of stimulus, but this interaction effect was not confirmed on the level of error rate. Furthermore, no differences in response were found between amputees who had lost their dominant or non-dominant limb. These findings are contrary to reports by (Nico et al., 2004) for upper limb amputees; they showed that upper limb loss significantly impaired performance in the same task, notably if the dominant limb was amputated.

Remarkably, the presence/absence of phantom sensation during testing did have an effect on the laterally judgments in our amputee group.

Evidence for the robustness of internal action representation comes from chronic hemiplegics (Johnson, 2000; Johnson et al., 2002). These patients preserved their ability to mentally simulate movements of the paralysed body part. Our results on a patient with rotationplasty illustrate that, even years after rotationplasty surgery, the patient retained the ability of unconstrained mental rotation. This is even more remarkable considering that the patient had adapted her motor behaviour. Rotationplasty patients, unlike amputees, receive sensory input from the periphery, i.e. their affected, but still present limb. As a consequence, we expected the patient to show a reversion in response time pattern for the rotated limb, if referring to her own body structure, since the post-surgical reorientation of her foot is different by 180° from a prototypical one. However, a slightly better performance for the affected limb in the plantar view can be seen. This trend may suggest some effect of the inversion of the foot in the rotationplasty patient; however, the effect is only a slight trend which may as well fit into the normal range. Ionta et al. (2007) proposed that the actual posture and the joined afferent information influence mental rotation of the same body part. However, the presented chronometric findings do not support this notion. The altered anatomical constraints are neither reflected in the response time nor in the error rate.

Berlucchi and Aglioti (1997) proposed that, despite the plasticity in the somatosensory cortex, the brain might be genetically predisposed to representing a prototypical human body. There is abundant evidence from neonatal research, as well as from phocomelic children for an innate origin of the body schema. In a landmark experiment, (Meltzoff and Moore, 1977) showed that neonates are capable of imitating orofacial and head movements of an adult model already days after delivery. This implies that they can equate their own and other people's body structure. Reports of phantom limb sensations in children with congenital limb deficiency provide further evidence for an innate body schema (Melzack et al., 1997). According to Melzack et al. (1997), the basic experience of the body is not a passive process that merely reflects sensory input from the body, but an interaction of genetic and sensory determinants. Evidence for multiple independent body representations comes from research by (Corradi-Dell'Acqua et al., 2009). They could show that the brain is endowed with two representations of the body maintained by different neural substrates: (1) the body schema, which represents the orientation of one's own body in space, and (2) the body structural description, representing the location of body parts relative to a prototypical body.

Laterality emerges very early in life and when dealing with motor imagery, laterality effects provide reliable evidence that subjects are referring to a representation of their own body, as found here in the group of healthy controls.

Accordingly, the disappearance of this functional marker represents a strong suggestion that the patients are not referring to a representation of their own body. Our findings suggest that the amputees and the patient with rotationplasty may be using a more abstract (prototypical) representation of the body, or more likely they may be operating mental visual transformations of the stimulus, thus not using motor imagery at all. It should be noted that mental rotation of a visual stimulus (rotating the picture of a foot, as opposed to imagining moving "my preferred foot") would entail exactly the disappearance of the laterality effect, while the effect of distance from a canonical orientation (i.e., a foot with the big toe pointing upward) would still be present. However, the elicitation of phantom sensations during the task in 2 of the 18 amputees suggests that at least these 2 patients were using motor imagery. This hypothesis should be tested in further studies, possibly making use of functional brain imaging during mental rotation tests in amputees.

Summary and Conclusions | 9

Summary

The series of studies presented in this thesis provide a neuromechanical perspective on movement in lower limb amputees, inquiring the complex interaction between *brain, body, prosthesis and environment* (Figure 1.1). Insights into the motor functioning and limitations in adaptability are gained through comparative analysis of able-bodied and amputee movement. This enables us to derive recommendations for prosthetic design and amputee rehabilitation.

In *Chapter 2* we characterized prosthetic foot properties by means of a newly-developed inverted pendulum-like apparatus. This device permits the emulation of a person's body weight acting on a prosthetic foot during roll-over. The testing of a broad range of prosthetic feet revealed highly distinct roll-over shapes. These roll-over shapes do not conform to an arc of constant radius. Shoes imposed small but functionally significant effects on the instantaneous radius of curvature, making it more constant.

A large radius of curvature is associated with stability. Therefore, amputees with poor balance will profit from a prosthetic foot with a large radius of curvature to improve standing stability.

Further investigations revealed that the roll-over shapes found in prosthetic feet approximated the "natural" roll-over shapes of able-bodied people (*Chapter 3*). However, despite this geometric similarity amputees did not walk symmetrically. Transfemoral amputees, in particular, showed highly individual adjustments in the roll-over shape on the side of the sound limb. Due to the passive properties of a prosthetic foot, the roll-over on this side is about the same in every step. Adjustments can thus only be made by the sound limb. In a complex *patient–prosthesis* interplay, amputees adjust the roll-over shapes of the sound limb to the limitations of the prosthesis. As the amount of stability provided by a prosthesis differs between models, it provides the opportunity to match a patient's balance capacity with the stability offered by her/his prosthesis.

To determine a person's balance capacity, a test for one-legged balance control was developed (*Chapter 4*). The special feature of this test is that its difficulty increases with the participant's balance capacity. By testing one-legged balance on ridges of gradually decreasing width, different balance control mechanisms are evoked. The narrow ridge balance test appeared to be a sensi-

tive tool, discriminating well between young and elderly controls. The test allows a fine-graded delimitation of balance control over a wide range, making it a promising tool for the assessment of one-legged balance control in lower limb amputees.

In *Chapter 5* we disentangled the relative contributions of the prosthetic and sound limb to balance control following waist-pull perturbations. It turned out that when being perturbed in anterior-posterior directions, amputees compensated for the absence of active ankle control in their prosthetic limb by increasing the ankle moment in their sound limb. In addition, the passive properties of the prosthetic foot contributed to balance control. In the medio-lateral perturbation condition, balance was regained by modulation of the lateral hip moments. As the hip joint remains intact after transtibial amputation, amputees experienced few limitations in this perturbation condition. When perturbations are large, other mechanisms come into play.

An impending fall (*Chapter 6*) can only be prevented by a protective stepping response. Transtibial amputees who were released from a forward-inclined orientation of 10% showed spatial and temporal differences when recovering using with their sound or their prosthetic limb first. When leading with their prosthetic limb, they responded faster and the interval between the heel-strike of the leading and trailing limb was shorter. Furthermore, amputees made a larger step and showed less knee flexion at heel-strike when leading with the prosthetic limb. Interestingly, the amputees as a group had no specific limb preference, prosthetic or sound, when recovering after a forward fall, despite the asymmetry of their locomotor system. Amputees appeared to be equally efficient in recovering from an impending fall as controls, irrespective of whether they led with their prosthetic or sound limb.

When the dynamic stability of prosthetic walking was challenged by irregularities of the ground (*Chapter 7*), no adjustments in the spatial and temporal stepping behavior were found. Instead of using a wider step width to increase lateral stability, the transtibial amputees amplified the lateral arm-swing velocity to increase stability on uneven ground.

For a better understanding of the changes in the mental representation of body and movement as a consequence of lower limb amputation and rotationplasty, a laterality recognition task was performed (*Chapter 8*). Study participants were shown illustrations of feet in different orientations, which they had to classify as left or right limb. This laterality recognition task is known to elicit implicit mental rotations of the subject's own body parts. The fact that the able-bodied controls showed the typical laterality effects, i.e. they recognized their preferred foot faster than their non-preferred foot, shows that it is a reliable marker for reference to their own body. The amputees and the rotationplasty patient, however, did not show such laterality effects, suggesting that they referred to a more prototypical body representation or performed mental visual transformation of the stimuli. Only two of the amputees experienced phantom sensations of their missing limb during the task. A possible interpretation is that they made use of *motor imagery*. Irrespectively, the overall findings suggest that the amputees used a different strategy in solving this task than the able-bodied controls.

Rethinking Prosthetic Design

> *"The human foot is a masterpiece of engineering and a work of art."*
> *– Leonardo da Vinci –*

The challenge in prosthetic engineering is to substitute the actions of the biological foot by means of a mechanical device. While the biological foot seamlessly adjusts to surface irregularities, walking speeds, gait initiation and termination, and stair decent and ascent, the roll-over properties of a prosthetic foot remain fixed. Prosthetic feet are typically made of springs, such as carbon leaf springs, and elastic bumpers. Mechanical testing has shown that the roll-over shapes of prosthetic feet agree closely with those created by biological feet in able-bodied walking, with the exception of the plantar flexion at push-off (*Chapter 2 & 3*). It appears that the equation for symmetric gait is a complex one. Although prosthetic feet have a "natural", able-bodied like roll-over shape, amputees show adjustments in their sound limb (*Chapter 3*). These adjustments were highly individual, just like a personal signature. When the prosthetic device is a passive system, all adjustments need to be made by the sound limb. Restoration of a symmetric gait pattern as a goal of rehabilitation should therefore be considered critically. Many of the adjustments may be regarded as functional to the passive device. It should be noted that although prosthetic and biological feet have geometrically similar roll-over shapes, the temporal dynamics of these shapes may not agree. Furthermore, roll-over shapes only account for the position of the CoP with respect to the angle of rotation of the limb but not for the direction of the ground reaction force.

Bionic feet have been developed in an effort to better mimic the function of the human ankle-foot system. Attempts have been made to improve efficiency of prosthetic walking by developing energy recycling feet (Collins and Kuo, 2010) as well as powered ankle-foot prostheses (Au et al., 2008; Eilenberg et al., 2010). To facilitate stair (Alimusaj et al., 2009) and ramp ambulation (Fradet et al., 2010) adaptive ankle-foot systems have been marketed. However, to date, such bionic feet are literally always a step behind, as only after having detected a change in environment, such as two steps in succession the ankle rotates to accommodate steps. This delay in transition between walking modes is also present in microprocessor-controlled prosthetic knees. Here, sensors and a microprocessor are used to regulate the stiffness of the knee, which for instance allows amputees to walk at different cadences. The transition between modes works fine for lasting changes. Yet, when a single stair or a short ramp needs to be negotiated the trigger is often too short to allow a transition between modes. While a biological limb allows a highly flexible immediate anticipatory response, a prosthetic limb, by contrast, needs time to detect and subsequently adjust to the changing demands. To avert this bottleneck myoelectric-driven ankle-foot prosthetics have been developed to allow a rapid transition between modes (Au et al., 2008). This prototypical prosthesis allows transtibial amputees to switch from level-ground walking to stair descent mode by contracting their gastrocnemius muscle during the swing phase before stepping onto the first stair. By contracting the tibialis anterior during the swing phase the ankle-foot prosthesis can be switched back to level-walking mode. This binary level control requires adequate myoelectric signals from the residual limb which may limit its applicability. Furthermore, powering the ankle joint goes along with an increase of weight of the prosthetic device due to the motors and the energy sources.

In prosthetic walking balance in the frontal plane is compromised due to the lack of the lateral ankle strategy resulting in a wider step width, increased stability margins (Hof et al., 2007) and lateral arm swing (*Chapter 7*). To tackle this deficit, a bar mechanism has been designed. Simulations revealed that this mechanism has the potential of enabling prosthetic walkers to actively modulate the direction of the ground reaction force by means of the hip moment of force during walking. Lateral stability is improved by enabling the amputee to generate more horizontal ground reaction forces (without tilting the limb) than with common prosthetic designs. Comprehensive prototype testing will have to show if the promising simulation outcomes hold true under real life conditions.

From a neuromechanical perspective an ideal prosthesis imposes the least disturbances of automated movements. This will minimize the amount of motor relearning required to adjust to the new dynamics of the motor system. For an optimal *patient-prosthesis* interplay the dynamic response properties of the new limb should match existing motor control strategies. Therefore a shift of the foremost technology-driven development of prosthetic devices towards a patient-driven approach could provide a fruitful ground for innovations.

Staying in Dynamic Balance

In walking the human body is perpetually losing and regaining balance (Hof et al., 2005), or as performance-artist Laurie Anderson (1982) describes it: we are "walking and falling at the same time". Dynamic stability of walking is achieved and maintained through a spatio-temporal adequate stepping response (Hof et al., 2005). After lower limb amputation patients need to adjust to the dynamics of the new artificial limb in order to regain safe and independent locomotion. Walking and standing with a leg prosthesis is challenged due to the passive properties of the artificial limb. In fact, the new limb lacks direct muscular control and sensory feedback, thereby hampering the stability of prosthetic walking and standing.

By mechanical testing it was shown that the roll-over properties of such feet vary widely between models but these differences diminish in combination with shoes (*Chapter 2*). The roll-over properties are directly linked to the stability provide by a prosthesis, i.e. the passive stability of the prosthetic foot is determined by its radius of curvature. For standing a radius of curvature greater than or equal to the height of the center of mass would result in a stable or metastable situation. Yet, prosthetic feet typically have a radius of curvature of about $^{1}/_{3}$ of the leg length, a curvature that was found to be metabolically most efficient for walking (Adamczyk et al., 2006). Effectively, the curvature of prosthetic feet is not constant; three segments can be identified, a begin and an end segment with a small instantaneous radius of curvature, and a middle section with a large instantaneous radius of curvature (*Chapter 2*). This segmentation nicely suits the different demands for standing and walking.

In the prosthesis fitting processes it is important to take the abilities of the patient and the properties of the prosthetic limb into account as the stability provided by the different prosthetic feet varies between models (*Chapter 2*). These passive properties of the prosthetic foot contribute to balance control and should be weighted against a patient's balance control and muscle strength (*Chapter 5*). One-legged balance controls has been identified as an important predictor for functional outcome after lower limb amputation (Schoppen et al., 2003). To allow a fine-graded delimitation of balance control, a one-legged balance control test was developed (*Chapter 4*), which may become a useful tool for the assessment of balance control in lower limb amputees.

In this thesis transtibial amputees and able-bodied controls faced complex motor tasks or challenging environments, such as walking on irregular surfaces (*Chapter 7*), balance recovery after an evoked fall (*Chapter 6*), and balance perturbations (*Chapter 5*) to gain insight into the functioning and limitations of the complex *brain, body, prosthesis and environment* interaction. Under these challenging conditions the transtibial amputees showed a variety of adjustment strategies to compensate for the lack of an active control of the prosthetic limb. For instance, they increased the contribution of the sound ankle to balance control when being perturbed in anterior-posterior direction (*Chapter 5*). Increasing the loading and the center of pressure displacement under the sound limb are common balance strategies in unilateral amputees, when standing on a platform performing translational movements in anterior-posterior direction (Vanicek et al., 2009b; Vrieling et al., 2008d). In mediolateral direction the hip strategy has been identified as a major balance strategy; as transtibial amputees have two fully functional, muscle empowered hip joints they showed no limitations in recovering balance after being perturbed in this direction (*Chapter 5*).

Remarkably the transtibial amputees were equally efficient in recovering balance after an evoked forward fall as able-bodied controls (*Chapter 6*), though the amputees did use slightly different strategies when leading with the sound versus the prosthetic limb. When leading with the prosthetic limb, they responded faster and took a longer step. Furthermore, the heel-strike interval was shorter and the knee flexion at heel strike was smaller. Taken as a whole ,these adjustments form an efficient strategy to break the forward fall when leading with the prosthetic limb first. Interestingly, the transtibial amputees as a group did not have a specific preference for recovering balance with the prosthetic or the sound limb. They recovered balance with the prosthetic limb just as well as with the sound limb leading. In other tasks, such as obstacle crossing, amputees appeared to have a limb preference, for example, transtibial amputees preferred to step over an obstacle with the prosthetic limb first, while transfemoral amputees preferred the sound limb. This difference in leading limb preference was attributed to the restricted flexion and propulsion properties of the prosthetic knee (Vrieling et al., 2007). It is likely that in the falls experiment transfemoral amputees would show a clear preference to recover balance with the sound limb leading as the prosthetic knee needs to be locked in order to bear weight. When descending stairs transtibial amputees use a step-over-step pattern (Powers et al., 1997). Transfemoral amputees with a mechanical knee, by contrast, make use of a step-by-step pattern. Yet, when having been fitted with a microprocessor-controlled prosthetic knee they more frequently show a step-over-step stair descent (Hafner et al., 2007). This illustrates that the presence or absence of the control of the knee is critical to the need for major movement adjustments.

During level treadmill walking transfemoral amputees were shown to compensate for the lack of lateral ankle strategy by increasing the stability margin for lateral balance on the prosthetic side (Hof et al., 2007). In a group of transtibial amputees walking on flat and irregular surfaces no such adjustments in lateral stability margins were found (*Chapter 7*). They were walking with a wide step, irrespective of the walking surface. Unlike the transtibial amputees, the able-bodied controls did increase their step width when walking on the irregular surface. Instead of adjusting the movements of the lower limbs transtibial amputees increased the lateral component of the arm-swing to maintain lateral walking stability on an irregular surface. This helped them to generate a greater horizontal ground reaction force component (Otten, 1999).

By focusing on the limitations of lower limb amputees facing complex motor tasks or challenging environments, valuable information on the nature of adaptation and its shortcomings can be obtained. These insights allow deducing targeted training methods and improvements of prosthetic design.

Clinical Implications

Amputee rehabilitation programs aim to enhance and restore independent locomotion. In order to function well outside an uncluttered training environment the amputee needs to regain movement flexibility with the prosthetic limb. To explore the possibilities of the movement repertoire represented in the neural system, challenging situations may be trained with a safety harness. The harness prevents the person from actually falling, while discovering what is possible with both the prosthetic and the sound limb. However, care should be taken that training with a "safety net" does not lead to a mental dependency.

It is remarkable that our experienced transtibial amputee participants performed quite well in recovering from the evoked forward fall, even with their prosthetic limb leading (*Chapter 6*). Several of them experienced this as a pleasant surprise, which added to their confidence. This experience suggests that amputees may benefit from bilateral fall training in rehabilitation, increasing their confidence in fall-prone situations. Yet, during the landing great impact forces act on the stump which bears the risk of injury. Given the rather critical review of microprocessor-controlled prosthetic knees (see above) it is only fair to note that they have been reported to reduce the frequency of stumbling and falling (Hafner et al., 2007).

In rehabilitation a stable and adaptive gait pattern should be promoted to minimize the risk of falling. After loss of a lower limb, amputees often desire to regain a symmetric gait pattern. However, the locomotor system of amputees is inherently asymmetric and all compensation movements need to take place in the sound limb. Therefore the focus of rehabilitation should be on training efficient strategies rather than gait symmetry. Furthermore, improving balance control and muscle strength will help to compensate for the passive properties of the prosthetic limb.

Future Perspectives

Attention should be given to the clinical implications of prosthetic research findings, so that the academic knowledge gained will indeed help improve individual well being. Thus, research should be truly translational in that it will move basic research findings to patient care, as well as, to the prosthetic industry. Currently a replication of the inverted pendulum-like apparatus (*Chapter 2*) is being used to evaluate the roll-over properties of prosthetic feet prototypes at the headquarters of Össur in Reykjavík, Iceland.

The narrow ridge balance test (*Chapter 4*) was shown to be a useful tool to measure balance over a wide spectrum of abilities. Follow-up research is warranted to investigate if the test can assist to determine an individual's optimal prosthesis composition. It is expected that individuals with poor balance control will profit from a more stable prosthesis, while those endowed with excellent balance control will depend less on the stability provided by the prosthesis. To optimize the prosthesis fitting process it will be important to further advance our understanding of the *patient–prosthesis* interplay. This can be achieved by manipulating the (roll-over) properties of prostheses in a systematic fashion and study their effects on walking and standing stability, while taking the motor abilities of the individual patient into account.

Finally, rehabilitation programs should aim to improve the *patient–prosthesis* interplay trough the training of complex movements. Challenging tasks such as described in this thesis (*Chapter 4–7*) should be implemented into amputee rehabilitation programs and the effectiveness of such training should be evaluated. Fewer fall incidences and an increase in mobility through more efficient training processes will also have multiple positive societal implications. It will allow amputees to play a more active role in society, be less dependent on medical care and last but not least help to reduce health care costs.

Recently the consortium SPRINT (Smart mobility devices with improved Patient pRosthesis INTeraction) has been initiated by the University Medical Center Groningen, the University of Groningen and the University of Twente. An active collaboration with large industries as well as small and medium enterprises is intended. SPRINT is recognized as one of eight centers of research excellence by the NWO-ZonMW (The Netherlands Organisation for Scientific Research – The Netherlands Organisation for Health Research and Development) in the IMDI.NL program (Innovative Medical Devices Initiative NL). An explicit focus of SPRINT is to develop the next generation of intelligent mobility devices for the individual patient. This will help to restore the mobility of patients and keep them independent for as long as possible and besides aims to decrease health care costs.

Conclusions

The studies subsumed in this thesis have added to our understanding of the complex interaction between *brain, body, prosthesis and environment*. We have developed tests that help to identify limits in prosthetic design and movement adaptation. We envisage that extensions of this work will lead to innovations in rehabilitation programs and patient-centered prosthesis prescription.

References

Adamczyk, P. G., Collins, S. H., Kuo, A. D., (2006). The advantages of a rolling foot in human walking. *Journal of Experimental Biology* 209, 3953–3963.

Alimusaj, M., Fradet, L., Braatz, F., Gerner, H. J., Wolf, S. I., (2009). Kinematics and kinetics with an adaptive ankle foot system during stair ambulation of transtibial amputees. *Gait & Posture* 30, 356–363.

Au, S., Berniker, M., Herr, H., (2008). Powered ankle-foot prosthesis to assist level-ground and stair-descent gaits. *Neural Networks* 21, 654–666.

Arampatzis, A., Karamanidis, K., Mademli, L., (2008). Deficits in the way to achieve balance related to mechanisms of dynamic stability control in the elderly. *Journal of Biomechanics* 41, 1754–1761.

Berg, K. O., Wood-Dauphinee, S. L., Williams, J. I., Maki, B., (1992). Measuring balance in the elderly: validation of an instrument. *Canadian Journal of Public Health* 83 Suppl 2, S7–11.

Berlucchi, G., Aglioti, S., (1997). The body in the brain: neural bases of corporeal awareness. *Trends in Neurosciences* 20, 560–564.

Beyaert, C., Grumillier, C., Martinet, N., Paysant, J., Andrq, J. M., (2008). Compensatory mechanism involving the knee joint of the intact limb during gait in unilateral below-knee amputees. *Gait & Posture* 28, 278–284.

Blakemore, S. J., Wolpert, D. M., Frith, C. D., (2002). Abnormalities in the awareness of action. *Trends in Cognitive Sciences* 6, 237–242.

Bonda, E., Petrides, M., Frey, S., Evans, A., (1995). Neural correlates of mental transformations of the body-in-space. *Proceedings of the National Academy of Sciences of the United States of America* 92, 11180–11184.

Brauer, S. G., Burns, Y. R., Galley, P., (2000). A prospective study of laboratory and clinical measures of postural stability to predict community-dwelling fallers. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences* 55, M469–M476.

Buxbaum, L. J., Coslett, H. B., (2001). Specialised structural descriptions for human body parts: Evidence from autotopagnosia. *Cognitive Neuropsychology* 18, 289–306.

Cohen, J., (1988). Statistical power analysis for the behavioral sciences. *Erlbaum*, New York.

Collins, S. H., Kuo, A. D., (2010). Recycling energy to restore impaired ankle function during human walking. *Public Library of Science ONE* 5, e9307.

Coren, S., (1993). The lateral preference inventory for measurement of handedness, footedness, eyedness, and earedness: norms for young adults. *Bulletin for the Psychonomic Society* 31, 1–3.

Corradi-Dell'Acqua, C., Tomasino, B., Fink, G. R., (2009). What is the position of an arm relative to the body? Neural correlates of body schema and body structural description. *Journal of Neuroscience* 29, 4162–4171.

Curtze, C., Hof, A. L., Otten, B., Postema, K., (2010a). Balance recovery after an evoked forward fall in unilateral transtibial amputees. *Gait & Posture* 32, 336–341.

Curtze, C., Hof, A. L., van Keeken, H. G., Halbertsma, J. P., Postema, K., Otten, B., (2009). Comparative roll-over analysis of prosthetic feet. *Journal of Biomechanics* 42, 1746–1753.

Curtze, C., Postema, K., Akkermans, H. W., Otten, E., Hof, A. L., (2010b). The Narrow Ridge Balance Test: A measure for one-leg lateral balance control. *Gait & Posture* 32, 627–631.

Cyr, M., Smeesters, C., (2007). Instructions limiting the number of steps do not affect the kinetics of the threshold of balance recovery in younger adults. *Journal of Biomechanics* 40, 2857–2864.

Cyr, M., Smeesters, C., (2009). Kinematics of the threshold of balance recovery are not affected by instructions limiting the number of steps in younger adults. *Gait & Posture* 29, 628–633.

de Lange, F. P., Hagoort, P., Toni, I., (2005). Neural topography and content of movement representations. *Journal of Cognitive Neuroscience* 17, 97–112.

de Lange, F. P., Helmich, R. C., Toni, I., (2006). Posture influences motor imagery: An fMRI study. *NeuroImage* 33, 609–617.

Do, M. C., Breniere, Y., Brenguier, P., (1982). A biomechanical study of balance recovery during the fall forward. *Journal of Biomechanics* 15, 933–939.

Donker, S. F., Beek, P. J., (2002). Interlimb coordination in prosthetic walking: effects of asymmetry and walking velocity. *Acta Psychologica* 110, 265–288.

Eilenberg, M. F., Geyer, H., Herr, H., (2010). Control of a powered ankle-foot prosthesis based on a neuromuscular model. *IEEE Transactions on Neural Systems and Rehabilitation Engineering : A Publication of the IEEE Engineering in Medicine and Biology Society* 18, 164–173.

Fradet, L., Alimusaj, M., Braatz, F., Wolf, S. I., (2010). Biomechanical analysis of ramp ambulation of transtibial amputees with an adaptive ankle foot system. *Gait & Posture* 32, 191–198.

Gaser, C., Schlaug, G., (2003). Brain structures differ between musicians and non-musicians. *Journal of Neuroscience* 23, 9240–9245.

Geil, M. D., (2002). An iterative method for viscoelastic modeling of prosthetic feet. *Journal of Biomechanics* 35, 1405–1410.

Geil, M. D., Parnianpour, M., Quesada, P., Berme, N., Simon, S., (2000). Comparison of methods for the calculation of energy storage and return in a dynamic elastic response prosthesis. *Journal of Biomechanics* 33, 1745–1750.

Geurts, A. C., Mulder, T. H., (1994). Attention demands in balance recovery following lower limb amputation. *Journal of Motor Behavior* 26, 162–170.

Geurts, A. C., Mulder, T. W., Nienhuis, B., Rijken, R. A., (1991). Dual-task assessment of reorganization of postural control in persons with lower limb amputation. *Archives of Physical Medicine and Rehabilitation* 72, 1059–1064.

Geurts, A. C., Mulder, T. W., Nienhuis, B., Rijken, R. A., (1992). Postural reorganization following lower limb amputation. Possible motor and sensory determinants of recovery. *Scandinavian Journal of Rehabilitation Medicine* 24, 83–90.

Grimer, R. J., (2005). Surgical options for children with osteosarcoma. The *Lancet Oncology* 6, 85–92.

Hafner, B. J., Willingham, L. L., Buell, N. C., Allyn, K. J., Smith, D. G., (2007). Evaluation of function, performance, and preference as transfemoral amputees transition from mechanical to microprocessor control of the prosthetic knee. *Archives of Physical Medicine and Rehabilitation* 88, 207–217.

Hansen, A. H., Childress, D. S., (2005). Effects of adding weight to the torso on roll-over characteristics of walking. *Journal of Rehabilitation Research & Development* 42, 381–390.

Hansen, A. H., Childress, D. S., (2004). Effects of shoe heel height on biologic rollover characteristics during walking. *Journal of Rehabilitation Research & Development* 41, 547–554.

Hansen, A. H., Childress, D. S., Knox, E. H., (2000). Prosthetic foot roll-over shapes with implications for alignment of transtibial prostheses. *Prosthetics and Orthotics International* 24, 205–215.

Hansen, A. H., Childress, D. S., Knox, E. H., (2004a). Roll-over shapes of human locomotor systems: effects of walking speed. *Clinical Biomechanics* 19, 407–414.

Hansen, A. H., Meier, M. R., Sam, M., Childress, D. S., Edwards, M. L., (2003). Alignment of transtibial prostheses based on roll-over shape principles. *Prosthetics and Orthotics International* 27, 89–99.

Hansen, A. H., Meier, M. R., Sessoms, P. H., Childress, D. S., (2006). The effects of prosthetic foot roll-over shape arc length on the gait of transtibial prosthesis users. *Prosthetics and Orthotics International* 30, 286–299.

Hansen, A. H., Sam, M., Childress, D. S., (2004b). The effective foot length ratio: a potential tool for characterization and evaluation of prosthetic feet. *Journal of Prosthetics and Orthotics* 16, 41–45.

Head, H., Holmes, G., (1911). Sensory disturbances from cerebral lesions. *Brain* 34, 102–254.

Hilliard, M. J., Martinez, K. M., Janssen, I., Edwards, B., Mille, M. L., Zhang, Y., Rogers, M. W., (2008). Lateral balance factors predict future falls in community-living older adults. *Archives of Physical Medicine and Rehabilitation* 89, 1708–1713.

Hof, A. L., (2005). Comparison of three methods to estimate the center of mass during balance assessment. *Journal of Biomechanics* 38, 2134–2135.

Hof, A. L., (2007). The equations of motion for a standing human reveal three mechanisms for balance. *Journal of Biomechanics* 40, 451–457.

Hof, A. L., (1996). Scaling gait data to body size. *Gait & Posture* 4, 222–223.

Hof, A. L., (2008). The "extrapolated center of mass" concept suggests a simple control of balance in walking. *Human Movement Science* 27, 112–125.

Hof, A. L., Gazendam, M. G. J., Sinke, W. E., (2005). The condition for dynamic stability. *Journal of Biomechanics* 38, 1–8.

Hof, A. L., van Bockel, R. M., Schoppen, T., Postema, K., (2007). Control of lateral balance in walking. Experimental findings in normal subjects and above-knee amputees. *Gait & Posture* 25, 250–258.

Hoogvliet, P., van Duyl, W., de Bakker, J., Mulder, P., Stam, H., (1997). A model for the relation between the displacement of the ankle and the center of pressure in the frontal plane, during one-leg stance. *Gait & Posture* 6, 39–49.

Horak, F. B., Nashner, L. M., (1986). Central programming of postural movements: adaptation to altered support-surface configurations. *Journal of Neurophysiology* 55, 1369–1381.

Hsiao-Wecksler, E. T., (2008). Biomechanical and age-related differences in balance recovery using the tether-release method. *Journal of Electromyography and Kinesiology* 18, 179–187.

Hsiao-Wecksler, E. T., Katdare, K., Matson, J., Liu, W., Lipsitz, L. A., Collins, J. J., (2003). Predicting the dynamic postural control response from quiet-stance behavior in elderly adults. *Journal of Biomechanics* 36, 1327–1333.

Ionta, S., Fourkas, A., Fiorio, M., Aglioti, S., (2007). The influence of hands posture on mental rotation of hands and feet. *Experimental Brain Research* 183, 1–7.

Johnson, S. H., (2000). Imagining the impossible: intact motor representations in hemiplegics. *NeuroReport* 11.

Johnson, S. H., Sprehn, G., Saykin, A. J., (2002). Intact motor imagery in chronic upper limb hemiplegics: Evidence for activity-independent action representations. *Journal of Cognitive Neuroscience* 14, 841–852.

Kendell, C., Lemaire, E. D., Dudek, N. L., Kofman, J., (2010). Indicators of dynamic stability in transtibial prosthesis users. *Gait & Posture* 31, 375–379.

King, D. L., Zatsiorsky, V. M., (2002). Periods of extreme ankle displacement during one-legged standing. *Gait & Posture* 15, 172–179.

Knecht, S., Henningsen, H., Elbert, T., Flor, H., Hohling, C., Pantev, C., Taub, E., (1996). Reorganizational and perceptional changes after amputation. *Brain* 119, 1213–1219.

Kuhtz-Buschbeck, J. P., Brockmann, K., Gilster, R., Koch, A., Stolze, H., (2008). Asymmetry of arm-swing not related to handedness. *Gait & Posture* 27, 447–454.

Lamoth, C. J., Ainsworth, E., Polomski, W., Houdijk, H., (2010). Variability and stability analysis of walking of transfemoral amputees. *Medical Engineering & Physics* 32, 1009–1014.

Lehmann, J. F., Price, R., Boswell-Bessette, S., Dralle, A., Questad, K., (1993). Comprehensive analysis of dynamic elastic response feet: Seattle Ankle/Lite Foot versus SACH foot. *Archives of Physical Medicine and Rehabilitation* 74, 853–861.

Lord, S. R., Rogers, M. W., Howland, A., Fitzpatrick, R., (1999). Lateral stability, sensorimotor function and falls in older people. *Journal of the American Geriatrics Society* 47, 1077–1081.

Luchies, C. W., Alexander, N. B., Schultz, A. B., Ashton-Miller, J., (1994). Stepping responses of young and old adults to postural disturbances: kinematics. *Journal of the American Geriatrics Society* 42, 506–512.

Maki, B. E., Holliday, P. J., Topper, A. K., (1994). A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population. *Journal of Gerontology* 49, M72–M84.

Massen, C. H., Kodde, L., (1979). A model for the description of left-right stabilograms. *Agressologie* 20, 107–108.

Mathias, S., Nayak, U. S., Isaacs, B., (1986). Balance in elderly patients: the "get-up and go" test. *Archives of Physical Medicine and Rehabilitation* 67, 387–389.

McGeer, T., (1990). Passive Dynamic Walking. *The International Journal of Robotics Research* 9, 62–82.

Meltzoff, A. N., Moore, M. K., (1977). Imitation of Facial and Manual Gestures by Human Neonates. *Science* 198, 75–78.

Melzack, R., (1992). Phantom Limbs. *Scientific American* 266, 120–125.

Melzack, R., Israel, R., Lacroix, R., Schultz, G., (1997). Phantom limbs in people with congenital limb deficiency or amputation in early childhood. *Brain* 120, 1603–1620.

Miff, S. C., Hansen, A. H., Childress, D. S., Gard, S. A., Meier, M. R., (2008). Roll-over shapes of the able-bodied knee-ankle-foot system during gait initiation, steady-state walking, and gait termination. *Gait & Posture* 27, 316–322.

Mille, M. L., Rogers, M. W., Martinez, K., Hedman, L. D., Johnson, M. E., Lord, S. R., Fitzpatrick, R. C., (2003). Thresholds for inducing protective stepping responses to external perturbations of human standing. *Journal of Neurophysiology* 90, 666–674.

Miller, W., Deathe, A. B., Speechley, M., Koval, J., (2001a). The influence of falling, fear of falling, and balance confidence on prosthetic mobility and social activity among individuals with a lower extremity amputation. *Archives of Physical Medicine and Rehabilitation* 82, 1238–1244.

Miller, W., Speechley, M., Deathe, B., (2001b). The prevalence and risk factors of falling and fear of falling among lower extremity amputees. *Archives of Physical Medicine and Rehabilitation* 82, 1031–1037.

Miller, W. C., Deathe, A. B., Speechley, M., (2003). Psychometric properties of the Activities-specific Balance Confidence Scale among individuals with a lower-limb amputation. *Archives of Physical Medicine and Rehabilitation* 84, 656–661.

Miller, W. C., Speechley, M., Deathe, A. B., (2002). Balance confidence among people with lower-limb amputations. *Physical Therapy* 82, 856–865.

Nederhand, M. J., Van Asseldonk, E. H., der Kooij, H. V., Rietman, H. S., (2011). Dynamic Balance Control (DBC) in lower leg amputee subjects;

contribution of the regulatory activity of the prosthesis side. *Clinical Biomechanics* doi:10.1016/j.clinbiomech.2011.07.008

Neptune, R. R., McGowan, C. P., (2011). Muscle contributions to whole-body sagittal plane angular momentum during walking. *Journal of Biomechanics* 44, 6–12.

Nico, D., Daprati, E., Rigal, F., Parsons, L., Sirigu, A., (2004). Left and right hand recognition in upper limb amputees. *Brain* 127, 120–132.

Otten, E., (2003). Inverse and forward dynamics: models of multi-body systems. *Philosophical Transactions of the Royal Society B: Biological Sciences* 358, 1493-1500.

Otten, E., (1999). Balancing on a narrow ridge: biomechanics and control. *Philosophical Transactions of the Royal Society B: Biological Science*s 354, 869–875.

Pai, Y. C., Rogers, M. W., Patton, J., Cain, T. D., Hanke, T. A., (1998). Static versus dynamic predictions of protective stepping following waist-pull perturbations in young and older adults. *Journal of Biomechanics* 31, 1111–1118.

Parsons, L. M., (1987). Imagined spatial transformations of one's hands and feet. *Cognitive Psychology* 19, 178–241.

Parsons, L. M., (1994). Temporal and kinematic properties of motor behavior reflected in mentally simulated action. *Journal of Experimental Psychology: Human Perception and Performance* 20, 709–730.

Parsons, L. M., Fox, P. T., Downs, J. H., Glass, T., Hirsch , T. B., Martin, C. C., Jerabek, P. A., Lancaster, J. L., (1995). Use of implicit motor imagery for visual shape discrimination as revealed by PET. *Nature* 375, 54–58.

Pick, A., (1922). Störung der Orientierung am eigenen Körper. *Psychological Research* 1, 303–318.

Pijnappels, M., Bobbert, M. F., van Dieen, J. H., (2004). Contribution of the support limb in control of angular momentum after tripping. *Journal of Biomechanics* 37, 1811–1818.

Pijnappels, M., Kingma, I., Wezenberg, D., Reurink, G., van Dieen, J. H., (2009). Armed against falls: the contribution of arm movements to balance recovery after tripping. *Experimental Brain Research* 201, 689–699.

Podsiadlo, D., Richardson, S., (1991). The timed "Up & Go": a test of basic functional mobility for frail elderly persons. *Journal of the American Geriatrics Society* 39, 142–148.

Postema, K., Hermens, H. J., de Vries, J., Koopman, H. F. J. M., Eisma, W. H., (1997). Energy storage and release of prosthetic feet Part 1: Biomechanical analysis related to user benefits. *Prosthetics and Orthotics International* 21, 17–27.

Powell, L. E., Myers, A. M., (1995). The Activities-specific Balance Confidence (ABC) Scale. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences* 50A, M28–M34.

Powers, C. M., Boyd, L. A., Torburn, L., Perry, J., (1997). Stair ambulation in persons with transtibial amputation: an analysis of the Seattle LightFoot. *Journal of Rehabilitation Research and Development* 34, 9–18.

Powers, C. M., Torburn, L., Perry, J., Ayyappa, E., (1994). Influence of prosthetic foot design on sound limb loading in adults with unilateral below-knee amputations. *Archives of Physical Medicine and Rehabilitation* 75, 825–829.

Prieto, T. E., Myklebust, J. B., Hoffmann, R. G., Lovett, E. G., Myklebust, B. M., (1996). Measures of postural steadiness: differences between healthy young and elderly adults. *IEEE Transactions on Biomedical Engineering* 43, 956–966.

Ramachandran, V. S., Rogers-Ramachandran, D., (2000). Phantom Limbs and Neural Plasticity. *Archives of Neurology* 57, 317–320.

Sadeghi, H., Allard, P., Duhaime, M., (1997). Functional gait asymmetry in able-bodied subjects. *Human Movement Science* 16, 243–258.

Schmid, M., Beltrami, G., Zambarbieri, D., Verni, G., (2005). Centre of pressure displacements in transfemoral amputees during gait. *Gait & Posture* 21, 255–262.

Schoppen, T., Boonstra, A., Groothoff, J. W., de Vries, J., Goeken, L. N., Eisma, W. H., (2003). Physical, mental, and social predictors of functional outcome in unilateral lower-limb amputees. *Archives of Physical Medicine and*

Rehabilitation 84, 803–811.

Schwoebel, J., Coslett, H. B., (2005). Evidence for Multiple, Distinct Representations of the Human Body. *Journal of Cognitive Neuroscience* 17, 543–553.

Schwoebel, J., Friedman, R., Duda, N., Coslett, H. B., (2001). Pain and the body schema: Evidence for peripheral effects on mental representations of movement. *Brain* 124, 2098–2104.

Scott, V., Votova, K., Scanlan, A., Close, J., (2007). Multifactorial and functional mobility assessment tools for fall risk among older adults in community, home-support, long-term and acute care settings. *Age and Ageing* 36, 130–139.

Silverman, A. K., Fey, N. P., Portillo, A., Walden, J. G., Bosker, G., Neptune, R. R., (2008). Compensatory mechanisms in below-knee amputee gait in response to increasing steady-state walking speeds. *Gait & Posture* 28, 602–609.

Sirigu, A., Grafman, J., Bressler, K., Sunderland, T., (1991). Multiple representations contribute to body knowledge processing: evidence from a case of autotopagnosia. *Brain* 114, 629–642.

Stel, V. S., Smit, J. H., Pluijm, S. M., Lips, P., (2003). Balance and mobility performance as treatable risk factors for recurrent falling in older persons. *Journal of Clinical Epidemiology* 56, 659–668.

Thelen, D. G., Wojcik, L. A., Schultz, A. B., Ashton-Miller, J. A., Alexander, N. B., (1997). Age differences in using a rapid step to regain balance during a forward fall. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences* 52, M8–13.

Thies, S. B., Richardson, J. K., Ashton-Miller, J. A., (2005). Effects of surface irregularity and lighting on step variability during gait: A study in healthy young and older women. *Gait & Posture* 22, 26–31.

Tinetti, M. E., (1986). Performance-oriented assessment of mobility problems in elderly patients. *Journal of the American Geriatrics Society* 34, 119–126.

Toh, S. L., Goh, J. C. H., Tan, P. H., Tay, T. E., (1993). Fatigue testing of energy storing prosthetic feet. *Prosthetics and Orthotics International* 17, 180–188.

van der Linden, M. L., Solomonidis, S. E., Spence, W. D., Li, N., Paul, J. P., (1999). A methodology for studying the effects of various types of prosthetic feet on the biomechanics of transfemoral amputee gait. *Journal of Biomechanics* 32, 877–889.

van Jaarsveld, H. W. L., Grootenboer, H. J., de Vries, J., Koopman, H. F. J. M., (1990). Stiffness and hysteresis properties of some prosthetic feet. *Prosthetics and Orthotics International* 14, 117–124.

van Nes, C. P., (1950). Rotationplasty for congenital defects of the femur. Making use of the ankle of the shortened limb to control the knee joint of a prosthesis. *The Journal of Bone and Joint Surgery* 32, 12–16.

Vanicek, N., Strike, S., McNaughton, L., Polman, R., (2009a). Gait patterns in transtibial amputee fallers vs. non-fallers: biomechanical differences during level walking. *Gait & Posture* 29, 415–420.

Vanicek, N., Strike, S., McNaughton, L., Polman, R., (2009b). Postural responses to dynamic perturbations in amputee fallers versus nonfallers: a comparative study with able-bodied subjects. *Archives of Physical Medicine and Rehabilitation* 90, 1018–1025.

Vrieling, A. H., van Keeken, H. G., Schoppen, T., Hof, A. L., Otten, B., Halbertsma, J. P., Postema, K., (2009). Gait adjustments in obstacle crossing, gait initiation and gait termination after a recent lower limb amputation. *Clinical Rehabilitation* 23, 659–671.

Vrieling, A. H., van Keeken, H. G., Schoppen, T., Otten, E., Halbertsma, J. P., Hof, A. L., Postema, K., (2007). Obstacle crossing in lower limb amputees. *Gait & Posture* 26, 587–594.

Vrieling, A. H., van Keeken, H. G., Schoppen, T., Otten, E., Halbertsma, J. P., Hof, A. L., Postema, K., (2008a). Gait initiation in lower limb amputees. *Gait & Posture* 27, 423–430.

Vrieling, A. H., van Keeken, H. G., Schoppen, T., Otten, E., Halbertsma, J. P., Hof, A. L., Postema, K., (2008b). Gait termination in lower limb amputees. *Gait & Posture* 27, 82–90.

Vrieling, A. H., van Keeken, H. G., Schoppen, T., Otten, E., Halbertsma, J. P., Hof, A. L., Postema, K., (2008c). Uphill and downhill walking in unilateral lower limb amputees. *Gait & Posture* 28, 235–242.

Vrieling, A. H., van Keeken, H. G., Schoppen, T., Otten, E., Hof, A. L., Halbertsma, J. P., Postema, K., (2008d). Balance control on a moving platform in unilateral lower limb amputees. *Gait & Posture* 28, 222–228.

Winter, D. A., (1995a). ABC of balance during standing and walking. *Waterloo Biomechanics*, Waterloo CA.

Winter, D. A., (1995b). Human balance and posture control during standing and walking. *Gait & Posture* 3, 193–214.

Winter, D. A., (1990). Biomechanics and motor control of human movement. *Wiley*, New York .

Winter, D. A., (1979). Biomechanics of Human Movement. *Wiley*, New York.

Winter, D. A., Prince, F., Frank, J. S., Powell, C., Zabjek, K. F., (1996). Unified theory regarding A/P and M/L balance in quiet stance. *Journal of Neurophysiology* 75, 2334-2343.

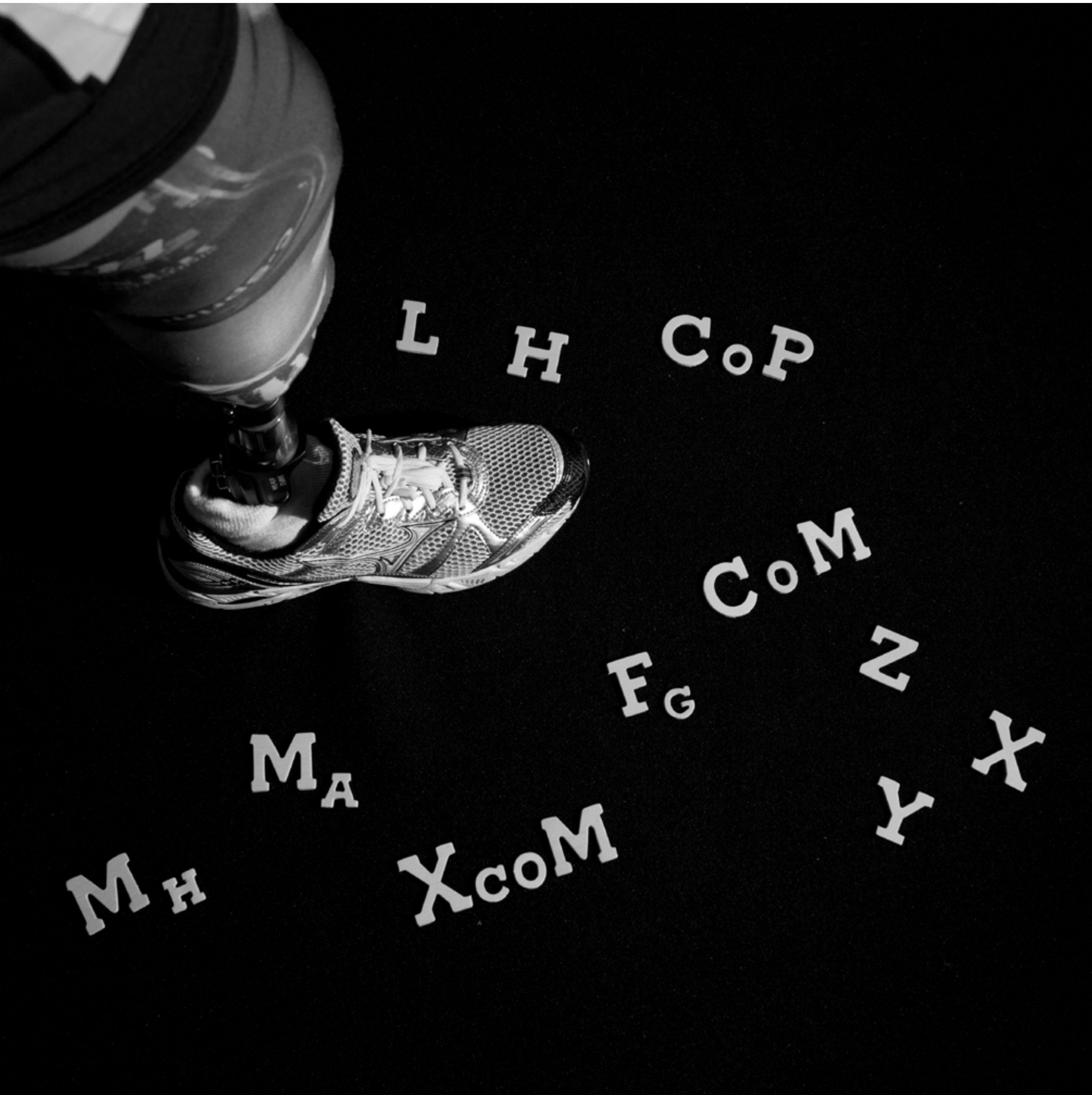
Wojcik, L. A., Thelen, D. G., Schultz, A. B., Ashton-Miller, J. A., Alexander, N. B., (1999). Age and gender differences in single-step recovery from a forward fall. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences* 54, M44–M50.

Woltering, H., (1985). On optimal smoothing and derivative estimation from noisy displacement data in biomechanics. *Human Movement Science* 4, 229–245.

Zmitrewicz, R. J., Neptune, R. R., Sasaki, K., (2007). Mechanical energetic contributions from individual muscles and elastic prosthetic feet during symmetric unilateral transtibial amputee walking: a theoretical study. *Journal of Biomechanics* 40, 1824–1831.

Zmitrewicz, R. J., Neptune, R. R., Walden, J. G., Rogers, W. E., Bosker, G. W., (2006). The effect of foot and ankle prosthetic components on braking and propulsive impulses during transtibial amputee gait. *Archives of Physical Medicine and Rehabilitation* 87, 1334–1339.

Nomenclature



$b_{\min}$	minimal distance between the anterior-posterior axis of the foot and the XcoM at foot contact
CoM	center of mass
CoP	center of pressure
F	force
F_{AM}	horizontal component of $\boldsymbol{F_G}$ generated by M_A
$\boldsymbol{F_G}$	ground reaction force
$\boldsymbol{F_{Gx}}$	forward component of $\boldsymbol{F_G}$
$\boldsymbol{F_{Gy}}$	lateral component of $\boldsymbol{F_G}$
$\boldsymbol{F_{Gz}}$	vertical component of $\boldsymbol{F_G}$
F_{HM}	horizontal component of $\boldsymbol{F_G}$ generated by M_H
g	acceleration of gravity (9.81 m/s)
H	height of CoM
l	effective pendulum length (trochanteric height times 1.24 and 1.34 for movements in the sagittal and frontal plane, respectively)
L	leg length (trochanteric height)
ln	natural logarithm
m	mass
M_A	ankle moment
M_{AP}	normalized ankle moment of the prosthetic limb
M_{AS}	normalized ankle moment of the sound limb
$\mathrm{M_H}$	hip moment
$\mathrm{M_{HP}}$	normalized hip moment of the prosthetic limb
M_{HS}	normalized hip moment of the sound limb
M_{sum}	sum of normalized moments
r.m.s.	root-mean square
s	forward travel

s'	instantaneous radius of curvature, i.e., first derivative of forward travel
v_{xCoM}	forward velocity of CoM
v_{yCoM}	lateral velocity of CoM
v_{yCoP}	lateral velocity of CoP
x, y, z	x-axis forward, y-axis medio-lateral to the left, z-axis vertically upward
x_0	center of curvature
x_{CoM}	forward position of CoM
XcoM	extrapolated center of mass
y_{CoM}	lateral position of CoM
y_{CoP}	lateral position of CoP
y_h	horizontal distance of CoM to the line of action of the ground reaction force; measure for the "counter-rotation" mechanism, proportional to arm and leg motion

Greek symbols

α	shank angle
ζ	lateral position of the XcoM
ξ	forward position of the XcoM
ρ	radius of curvature
ω_0	eigenfrequency of pendulum ($\sqrt{g/l}$)

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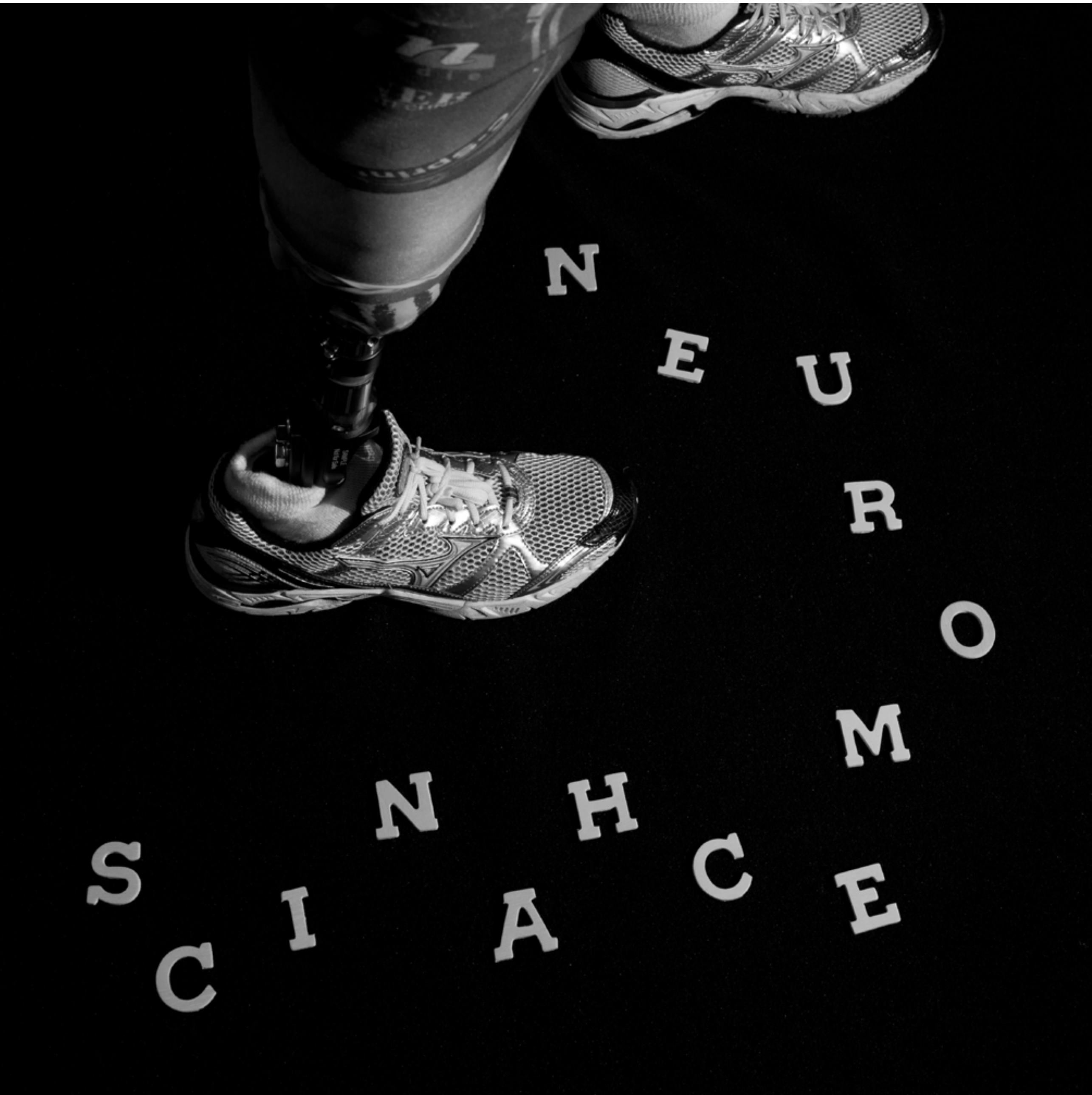

A

B

C

M

N

P

R

S

T

X

Samenvatting

Dutch Summary

Neuromechanica van het bewegen bij mensen met een beenamputatie

Dit proefschrift presenteert een neuromechanisch perspectief op het bewegen van mensen met een beenamputatie, en op de complexe interactie tussen *hersenen, lichaam, prothese en omgeving* (Figuur 1.1). Met een vergelijkende analyse van het bewegen bij gezonden en bij geamputeerden ontstaat inzicht in het motorisch functioneren en de grenzen van de mogelijkheden tot aanpassing. Hieruit volgen aanbevelingen voor protheseontwerp en revalidatieprogramma's.

In *Hoofdstuk 2* hebben we de eigenschappen van prothesevoeten onderzocht met behulp van een nieuw apparaat, een soort omgekeerde slinger. Dit apparaat simuleerde de belasting van het lichaamsgewicht die tijdens de afwikkeling op een prothesevoet werkt. Het testen van een groot scala prothesevoeten bracht duidelijk verschillende afwikkelprofielen aan het licht. Deze komen niet overeen met een cirkelboog van constante straal. Schoenen veroorzaakten kleine, maar functioneel significant effecten waardoor de kromtestraal constanter werd. Een grote kromtestraal gaat samen met stabiliteit. Daarom zullen motorisch minder vaardige geamputeerden voordeel hebben van een prothesevoet met een grote kromtestraal om de stabiliteit tijdens het staan te verbeteren.

Uit vervolgonderzoek bleek dat de afwikkelprofielen van prothesevoeten die van gezonde proefpersonen benaderden (*Hoofdstuk 3*). Ondanks deze geometrische overeenkomst, liepen geamputeerden echter niet symmetrisch. Met name bovenbeengeamputeerden lieten veel individuele aanpassingen zien in het afwikkelprofiel van het gezonde been. Als gevolg van de passieve eigenschappen van een prothesevoet is in het prothesebeen de afwikkeling in elke stap ongeveer dezelfde. Aanpassingen kunnen dus alleen gedaan worden door het gezonde been. In een complexe wisselwerking *patiënt-prothese* passen geamputeerden de afwikkeling van het gezonde been aan de beperkingen van de prothese aan. Aangezien de mate van stabiliteit die diverse protheses bieden verschilt, is het mogelijk de balansvaardigheid van de patiënt en de stabiliteit die een prothese biedt, aan elkaar aan te passen.

Om de balansvaardigheid te bepalen, is een test voor balanshandhaving op één been ontwikkeld (*Hoofdstuk 4*). Specifiek kenmerk van deze test is dat de moeilijkheidsgraad toeneemt met de balansvaardigheid van de proefpersoon. Door de balans op één been te testen op richels van gradueel afnemende breedte, komen verschillende balansmechanismen aan bod. Deze balanstest bleek een gevoelig instrument, dat goed discrimineerde tussen jonge en oudere proefpersonen. De test meet de kwaliteit van de balans met een hoog scheidend vermogen over een groot bereik, waardoor het een veelbelovend instrument is voor het bepalen van de balans op één been bij beengeamputeerden.

In *Hoofdstuk 5* hebben we onderscheid gemaakt in de bijdragen die het prothesebeen en het gezonde been hebben in de bij een balansverstoring op heuphoogte. Het bleek dat bij verstoring in voor-achterwaartse richting geamputeerden de afwezigheid van actieve enkelsturing in het prothesebeen compenseerden met een toegenomen enkelmoment in het gezonde been. Ook de passieve eigenschappen van de prothesevoet droegen bij aan de balanshandhaving. Bij medio-laterale verstoring werd de balans hersteld door aanpassing van het laterale heupmoment. Aangezien het heupgewricht intact blijft na transtibiale amputatie, ondervinden deze geamputeerden weinig beperkingen bij verstoring van het evenwicht in medio-laterale richting. Bij grote verstoringen spelen andere factoren een rol.

Een dreigende val (*Hoofdstuk 6*) kan alleen worden voorkomen door een beschermende stapreactie. Onderbeengeamputeerden, die werden losgelaten bij 10% naar voren leunen, lieten spatiotemporele verschillen zien bij het herstellen van hun evenwicht, al naar gelang ze uitstapten met het gezonde been, dan wel met het prothesebeen. Als ze uitstapten met het prothesebeen, reageerden ze sneller en het interval tussen hielcontact van het uitstappende en het volgende been was korter. Ook maakten geamputeerden een grotere stap en toonden ze minder knieflexie bij hielcontact als het prothesebeen leidde. Opmerkelijk is dat geamputeerden als groep geen voorkeursbeen hadden om te herstellen van een val voorwaarts, ondanks de asymmetrie van hun bewegingsapparaat. Geamputeerden leken even efficiënt in het herstellen van een dreigende val als de controlegroep, ongeacht of ze uitstapten met het prothesebeen dan wel het gezonde been.

Als de dynamische stabiliteit van het lopen met een prothese bemoeilijkt werd door oneffenheden in het oppervlak (*Hoofdstuk 7*), werden geen aanpassingen in het spatiotemporele looppatroon gevonden. In plaats van een grotere stapbreedte te kiezen om de zijwaartse stabiliteit te vergroten, vergrootten de onderbeengeamputeerden de zijwaartse snelheid van de armzwaai om stabiel te lopen op een oneffen oppervlak.

Om een beter begrip van de veranderingen in de mentale representatie van lichaam en beweging als gevolg van beenamputatie en omkeerplastiek te verkrijgen, werd een links-rechts herkenningstaak uitgevoerd (*Hoofdstuk 8*). Proefpersonen kregen illustraties te zien van voeten in verschillende standen die zij moesten classificeren als linker of rechter voet. Van deze taak is bekend dat hij impliciete mentale rotaties van het eigen lichaamsdeel uitlokt. Het feit dat de controlegroep typische lateraliteitseffecten liet zien, met name dat zij hun voorkeursvoet sneller herkenden dan de niet-voorkeursvoet, impliceert dat het een betrouwbare indicatie is voor betrekking op het eigen lichaam. De geamputeerden en de omkeerplastiek-patiënt vertoonden deze lateraliteitseffecten echter niet, wat suggereert dat zij gebruikmaakten van andere strategieën, zoals een meer prototypisch lichaamsbeeld of ruimtelijke rotatie van de stimuli. Slechts twee geamputeerden ervoeren fantoomsensaties in hun afwezige lichaamsdeel tijdens de taak. Het is mogelijk dat zij gebruikmaakten van de strategie *motor imagery*. De algemene bevindingen suggereren hoe dan ook dat geamputeerden gebruikmaakten van een andere strategie voor deze taak dan de personen in de controlegroep.

Zusammenfassung

German Summary

Neuromechanik des Bewegens bei Beinamputierten

Die in dieser Dissertation präsentierte Forschungsreihe gibt eine neuromechanische Perspektive des Bewegens bei Beinamputierten; ergründet wird die komplexe Interaktion von *Gehirn, Körper, Prothese und Umgebung* (Abbildung 1.1). Durch die vergleichende Analyse der Bewegungen von Gesunden und Amputierten wurden Erkenntnisse über das motorische Funktionieren und die Anpassungsgrenzen gewonnen. Dies ermöglicht es uns, Empfehlungen für Prothesendesign und Rehabilitation von Amputierten abzuleiten.

In *Kapitel 2* untersuchten wir die Eigenschaften von Prothesenfüßen mittels eines neu entwickelten Apparates, einer Art inverses Pendel. Dieser Apparat simuliert die Gewichtsbelastungen, die während des Abrollens auf einen Prothesenfuß wirken. Das Testen einer Vielzahl an Prothesenfüßen ergab stark unterschiedliche Abrollkurven. Diese Abrollkurven entsprechen keinem Kreis mit konstantem Radius. Schuhe beeinflussen die Abrollkurven nur leicht, jedoch funktionell signifikant; sie erzeugen einen konstanteren Radius. Ein großer Radius ist gleichbedeutend mit mehr Stabilität. Daher profitieren Amputierte mit schlechtem Gleichgewicht von einem Prothesenfuß mit großem Radius, welcher ihre Stabilität beim Stehen verbessert.

Weitere Untersuchungen haben ergeben, dass die Abrollkurven von Prothesenfüßen denen von gesunden Personen ähneln (*Kapitel 3*). Trotz dieser geometrischen Überseinstimmung gehen Amputierte jedoch nicht symmetrisch. Vor allem Oberschenkelamputierte zeigten höchst individuelle Anpassungen in der Abrollkurve des gesunden Beines. Aufgrund der passiven Eigenschaften von Prothesenfüßen ist die Abrollkurve bei jedem Schritt dieselbe, wohingegen das gesunde Bein Anpassungen machen kann. In einem komplexen Wechselspiel zwischen *Patient–Prothese*, passen Amputierte die Abrollkurve des gesunden Beines an die Einschränkungen der Prothese an. Da die durch eine Prothese gegebene Stabilität zwischen den Modellen variiert, müssen die Gleichgewichtsfähigkeit eines Patienten und die durch die Prothese gebotene Stabilität einander komplementieren.

Um die Gleichgewichtsfähigkeit von Personen zu bestimmen, wurde ein Test zur Messung der einbeinigen Balancekontrolle entwickelt (*Kapitel 4*). Eine besondere Eigenschaft dieses Tests ist, dass die Schwierigkeit mit der Gleichgewichtsfähigkeit der Versuchsperson zunimmt. Durch das Testen des Einbeinstands auf T-Schienen von graduell abnehmender Breite werden verschiedene Balancemechanismen abgerufen. Der Balancetest erwies sich als ein sensitives Messinstrument, das gut zwischen jungen und älteren Versuchspersonen diskriminiert. Der Test erlaubt eine feinstufige Abgrenzung der Balancekontrolle über einen weiten Bereich, was den Test zu einem vielversprechenden Instrument zur Bestimmung der Balance im Einbeinstand bei Beinamputierten macht.

In *Kapitel 5* haben wir die relativen Beiträge des Prothesenbeines und des gesunden Beines zur Wiederherstellung der Balance nach einer externen Störung der Balance auf Hüfthöhe bestimmt. Bei externer Störung der Balance in anterior-posterior Richtung kompensierten Amputierte die fehlende aktive Kontrolle des Fußgelenkes im Prothesenbein, indem sie den Moment im Fußgelenk des gesunden Beines erhöhten. Darüber hinaus trugen auch die passiven Eigenschaften des Prothesenfußes zur Balancekontrolle bei. Bei externer Störung der Balance in medio-lateraler Richtung wurde die Balance durch Modulation der lateralen Hüftmomente wieder hergestellt. Angesichts der Tatsache, dass das Hüftgelenk nach einer Unterschenkelamputation intakt bleibt, erfuhren diese Beinamputierten nur wenige Einschränkungen bei externen Störungen der Balance in medio-lateraler Richtung. Bei großen externen Störungen der Balance spielen andere Mechanismen eine Rolle.

Ein drohender Fall (*Kapitel 6*) kann nur durch eine schützende Schrittreaktion abgewendet werden. Unterschenkelamputierte, die aus einer nach vorn geneigten Lage von 10% losgelassen wurden, zeigten räumliche und zeitliche Unterschiede, je nachdem ob sie sich mit dem Prothesenbein oder dem gesunden Bein abfingen. Wenn sie sich mit dem Prothesenbein abfingen, reagierten sie schneller und das Intervall zwischen den Fersenkontakten des führenden und des folgenden Beines war kürzer. Darüber hinaus vergrößerten die Amputierten ihre Schrittlänge und zeigten eine geringere Flexion des Kniegelenkes zum Zeitpunkt des Fersenkontaktes, wenn sie sich mit dem Prothesenbein abfingen. Bemerkenswert ist, dass die Amputierten als Gruppe betrachtet keine Präferenz hatten, sich mit dem Prothesenbein oder dem gesunden Bein abzufangen, trotz der Asymmetrie ihres Bewegungsapparates. Die Amputierten waren gleichermaßen effizient einen drohender Fall abzuwenden wie die Kontrollgruppe, unabhängig davon, ob sie sich mit dem Prothesenbein oder dem gesunden Bein abfingen.

Wenn die dynamische Balance des Gehens mit Prothese durch Unregelmäßigkeiten des Untergrundes erschwert wurde, konnte keine Veränderungen des raum-zeitlichen Gangbildes festgestellt werden (*Kapitel 7*). Anstatt die Gangspur zu verbreitern, um die laterale Stabilität zu erhöhen, vergrößerten Unterschenkelamputierte die Geschwindigkeit des lateralen Armschwungs, um sicher auf unebenem Untergrund zu gehen.

Zum besseren Verständnis der Veränderungen in der mentalen Repräsentation von Körper und Bewegung als Folge von Beinamputation und Umkehrplastik, wurde eine mentale Rotationsaufgabe durchgeführt (*Kapitel 8*). Den Versuchspersonen wurden Abbildungen von Füßen in verschiedenen Orientierungen gezeigt, welche sie als linken oder rechten Fuß klassifizieren mussten. Von dieser Aufgabe ist bekannt, dass sie implizit die mentale Rotation des eigenen Körperteiles hervorruft. Der Befund, dass die Kontrollgruppe typische Lateralitätseffekte zeigte, d.h. sie erkannten ihren dominanten Fuß schneller als den Anderen, ist ein zuverlässiger Indikator dafür, dass sie sich auf ihren eigenen Körper bezogen. Die Amputierten und die Patienten mit Umkehrplastik hingegen zeigten diesen Lateralitätseffekt nicht, was nahelegt, dass sie sich auf ein prototypisches Körperbild bezogen oder mentale Rotation des visuellen Objektes durchführten. Lediglich zwei der Amputierten erfuhren während der Aufgabe Phantomgefühle des amputierten Körperteiles. Eine mögliche Interpretation ist, dass sie mentale Bewegungsvorstellung (*motor imagery*) zur Lösung der Aufgabe anwandten. Ungeachtet dessen, zeigen die Ergebnisse insgesamt, dass die Amputierten eine andere Strategie zur Lösung der Aufgabe anwandten als die Personen in der Kontrollgruppe.

Acknowledgments

Acknowledgments

This thesis is a culmination of many people's contributions to whom I am truly grateful.

Mijn eerste promotor, professor dr. K. Postema. Beste Klaas, geweldig bedankt voor de geboden kansen, de vrijheid, het enorme vertrouwen dat je in mij stelde. Ik heb veel geleerd van jouw kijk op de klinische betekenis van resultaten, en de link die je steeds weet te leggen met de praktijk. Je enthousiasme — "Wat prachtig!" — werkte aanstekelijk.

Mijn tweede promotor, professor dr. E. Otten. Beste Bert, met je voortdurend nieuwe ideeën en je kritische kijk op bestaande ideeën was je voor mij een enorme bron van inspiratie. Je was niet alleen als wetenschapper maar ook als fotograaf betrokken bij dit promotietraject. Jouw fotowerk laat zien hoe kunst en wetenschap kunnen versmelten.

Mijn co-promotor, dr. ir. A. L. Hof. Beste At, als *Doktorvater* maak jij mijn begeleidingstrojka compleet. Met je enorme kennis en ervaring heb je een essentiële rol gespeeld in het slagen van dit project. Ik heb genoten van de vrijdagmiddag-lunches, de gesprekken over wetenschap en het leven in het algemeen. Je bent een wetenschapper *pur sang* en ook na je officiële afscheid bleef jij nog even sterk betrokken bij dit onderzoek.

Dr. A. H. Vrieling en drs. H. G. van Keeken, de voortrekkers van mijn onderzoek. Aline en Helco, het was een eer om op jullie ervaring verder te mogen bouwen. Helco, jouw Matlab Crash-cours was de basis voor vele ontdekkingen, beschreven in dit proefschrift.

Dr. J. P. K. Halbertsma. Jan, hartelijk dank voor je adviezen in de startfase van het onderzoek.

Professor dr. P. U. Dijkstra en dr. R. E. Stewart. Beste Pieter en Roy, ik heb veel geleerd van onze statistische discussies, waarvoor mijn dank en waardering.

Ronald Davidsz: jij hebt de puntjes op de *i* gezet, ofwel het goede label aan de juiste marker toegevoegd. Jouw computerkennis was een zegen voor dit project. Voor je enorme inspanningen bij alle metingen: ontzettend bedankt.

Henk Zijlstra: jouw enthousiasme, enorme kennis en gevoel voor het vak, technische steun en adviezen waren zeer waardevol. Ik ben blij dat ik wat heb kunnen terugdoen in de afronding van je bachelor-fase. Behalve van onze samenwerking, heb ik ook erg genoten van onze tennispartijtjes.

Beste vrijwilligers, amputatiepatiënten en andere proefpersonen. Zonder jullie was dit onderzoek er helemaal niet geweest. Zeer bedankt voor jullie bereidheid aan de experimenten mee te werken. Ik heb van jullie niet alleen geleerd hoe je op een prothese loopt, maar vooral hoe je met een prothese leeft. Beste Jacco, hartelijk dank dat jij zelfs nog eens terug wilde komen om als fotomodel te fungeren.

Dank ook aan iedereen die meedacht: Bert Slootmaker van de Instrumentmakerij voor zijn creatief ontwerp van het omgekeerde slingerapparaat; Cojanne Kars voor haar krachtige ondersteuning bij het meten van prothesevoeten; Hilda Akkermans, mijn eerste afstudeerder, voor de samenwerking bij het balanceren op richels; Merce Salvans-Batrons for her assistance with the fabrication of the "struikeltapijt"; Sjoerd de Vries en Emyl Smid voor hun steun bij het opzetten van de Parson's taak; de medewerkers van het secretariaat en de zorgadministratie voor hun organisatorische steun; Maerian de Jong, mijn persoonlijke Nederlandse *ghostwriter*, voor de redactionele bijdragen.

Mijn collega's Aline Vrieling, Bianca Nijhuis, Corine van Es, Eva Jaarsma, Ecatharina Golea Vasluian, Gerda Reinsma, Henk Meulenbelt, Ingrid de Jong, Jaap van Netten, Jesse Verschuren, Juha Hijmans, Lauren Fortington, Lex de Jong, Lonneke Opsteegh, Marlies Bodde, Martin Stenekes, Mihai Bragaru, Sietske Romkema, Sobhan Sobhani en Wietske Kuijer: bedankt voor jullie niet-aflatende steun en betrokkenheid en de gezelligheid. Mocht iemand denken dat onderzoek doen eenzaam werk is: niet in Groningen!

My sincere gratitude goes to the members of the reading committee professor dr. B. Drerup, professor dr. J. E. J. Duysens and professor dr. ir. H. F. J. M. Koopman for evaluating my thesis.

Professor Dr. J. Freiwald möchte ich dafür danken, dass er mir den Weg nach Groningen gewiesen hat.

Na vele uren achter de computer waren er de eureka-momenten bij het sporten. Vrienden van UMCG - Fitness: bedankt voor het sportplezier en de ontspanning. Een dag niet gesport, is een dag niet geleefd!

Anna Muijzer: we zijn samen naar de finish gegaan - ik heb veel gehad aan onze (altijd gezellige) gesprekken. Volgende week is het jouw beurt.

Professor Dr. R. Breitling. Lieber Rainer, many thanks for the countless hours of mentoring in which you shared your manifold expertise, for letting me bounce ideas off you and for joining me on all the late night cups of tea.

My two and a half paranymphs, dear Ingrid, Bettina and Katja, many thanks for your great support on this special day. Ingrid en Katja, bedankt voor jullie meeleven, meedenken, en oprechte interesse tijdens mijn Groningse tijd. Bettina, gemeinsam haben wir schon so manches Abenteuer durchlebt und ich bin froh, auch bei diesem Abenteuer meine große Schwester an meiner Seite zu wissen. Ladies, let's bring it on!

Lieve Bärbel, je zag het al helemaal voor je: Tijs en Jop met een strikje om. Maar jullie plekken op de eerste rij blijven vandaag leeg. In Olivier en Jop zijn jij en Tijs er toch nog bij. Jullie waren mijn tweede familie, jullie namen mij op in het gezin en leefden mee met ups en downs, met frustratie en blijdschap. Ik dank je voor je enorme betrokkenheid en diepe vriendschap. Ik mis jullie elke dag.

Liebe Eltern, vielen Dank für Eure uneingeschränkte Unterstützung, Ermutigung und Interesse. Papa, Dein tatkräftiger Einsatz bei der Herstellung der erforderlichen Messapparatur war großartig.

Haren, 7 March 2012

Carolin

About the Author

Curriculum Vitae

Carolin Curtze (1979) is a PhD student at the University Medical Center Groningen, the Netherlands, where in 2005 she started her PhD project on the "Neuromechanics of movement in lower limb amputees". Prior to that, she studied at the University of Gießen, Germany, where in 2003, she obtained her First State Examination for Primary School Teachers and then continued on with a master's degree in Sports Science with minors in Psychology and German. During her studies she worked as a research assistant in joint projects of the University of Greifswald and the University of Gießen on motor learning and control. Upon graduation, Carolin is looking to continue her research in the field of Neuromechanics.

Peer-reviewed Journals

Curtze, C., Otten, E., Hof, A. L., Postema, K., (2011). Determining asymmetry of roll-over shapes in prosthetic walking. *Journal of Rehabilitation Research & Development* 48(10), 1249–1260.

Curtze, C., Hof, A. L., Postema, K., Otten, E., (2011). Over rough and smooth: Amputee gait on an irregular surface. *Gait & Posture* 33(2), 292–296.

Curtze, C., Postema, K., Akkermans, H. W., Otten, E., Hof, A. L., (2010). The Narrow Ridge Balance Test: A measure for one-leg lateral balance control. *Gait & Posture* 32(4), 627–631.

Curtze, C., Hof, A. L., Otten, E., Postema, K., (2010). Balance recovery after an evoked forward fall in unilateral transtibial amputees. *Gait & Posture* 32(3), 336–341.

Curtze, C., Otten, E., Postema, K., (2010). Effects of lower limb amputation on the mental rotation of feet. *Experimental Brain Research* 201(3), 527–534.

Curtze, C., Hof, A. L., van Keeken, H. G., Halbertsma, J. P. K., Postema, K., Otten, E., (2009). Comparative roll-over analysis of prosthetic feet. *Journal of Biomechanics* 42(11), 1746–1753.

Conference Contributions

Curtze, C., Hof, A. L., Otten, E., Postema, K., (2010). Balance recovery after a simulated fall in lower limb amputees. *13th World Congress of the International Society for Prosthetics and Orthotics*, May 12–15, Leipzig, Germany.

Curtze, C., Otten, E., Hof, A. L., Postema, K., (2010). Asymmetry of roll-over in prosthetic walking. *13th World Congress of the International Society for Prosthetics and Orthotics*, May 12–15, Leipzig, Germany.

Curtze, C., Otten, E., Postema, K., (2010). Effects of lower limb amputation on the mental rotation of feet: an analysis of constraints and plasticity. *Neural Control of Movement*, April 20–25, Naples Florida, US.

Curtze, C., Hof, A. L., van Keeken, H. G., Halbertsma, J. P. K., Otten, E., Postema, K., (2009). Lower limb amputation and the ability to reorganize motor control. *European Society of Movement Analysis in Adults and Children*, September 14–19, London, UK.

Curtze, C., Hof, A. L., van Keeken, H. G., Halbertsma, J. P. K., Postema, K., Otten, E. (2009). Comparative biomechanics of prosthetic feet. *European Society of Movement Analysis in Adults and Children*, September 14–19, London, UK. (oral presentation)

Curtze, C., Otten, E., Postema, K., (2009). Mental representation of leg movement in lower-limb amputees: constraints or plasticity? *Progress in Motor Control VII*, July 23–25, Marseille, France.

Curtze, C., Reiser, M., (2003). Einfluss unterschiedlicher Laufgeschwindigkeiten auf die Bewegungsmuster bei Freizeit- und Wettkampfläufern. p. 308. In Strauß, B., Hagemann, N., Tietje, M. & G. Falkenberg-Gurges (eds.), *sport goes media — abstracts*. Hamburg: Czwalina.

Invited Talks

Symposium: Electromyografie – Elasticiteit – Evenwicht, Groningen, The Netherlands, January 27, 2011

Klinik und Poliklinik für Technische Orthopädie und Rehabilitation, Universitätsklinikum Münster, Germany, July 3, 2007

8e Symposium Revalidatietechniek, Groningen—Enschede, The Netherlands, May 19, 2006

Institut für Sportwissenschaft, University of Gießen, Germany, December 13, 2005

Colophon

Design	Carolin Curtze www.ccurtze.com
Photography	Bert Otten www.kalons.nl (except page 21)
Printer	Gildeprint Drukkerijen www.gildeprint.nl
Cover	240 gram
Book Block	115 gram g-print
Color	Pantone Black C
Fonts	Minion Pro, Myriad Pro

Financial Support

The research of this thesis was financially supported by:

- OIM Foundation
- Stichting Beatrixoord Noord-Nederland

Printing of this thesis was financially supported by:

- Centrum voor Revalidatie, Universitair Medisch Centrum Groningen
- D.H. Heijne Stichting / Basko Healthcare
- Graduate School of Medical Sciences, University Medical Center Groningen
- OIM Orthopedie
- Stichting Beatrixoord Noord-Nederland